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# Effects of Wider Step Width on Knee Biomechanics in Obese and Healthy-Weight Participants During Stair Ascent

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# Abstract

An increased likelihood of developing obesity-related knee osteoarthritis may be associated with increased peak internal knee abduction moment. Increases in step width may act to reduce this moment. This study focused on how step width influenced the knee joint during stair ascent by healthy and obese participants. Participants ascended stairs while walking at their preferred speed and under one of two step width conditions – preferred and increased. Obese participants experienced greater mediolateral and vertical ground reaction forces (GRFs), as well as increased peak knee extensor moments and push-off peak internal knee adduction moments. The findings of this study indicate that when step width increases, obese participants will experience a disproportionate increase in Loading-response and push-off response peak mediolateral GRF, push-off peak knee adduction moments, and peak knee adduction angle compared to healthy participants. When normalized to lean body mass, obese participants also had greater increases in peak knee extension moments under the increased step width condition. Participants in each group experienced decreased in loading-response peak vertical GRF, loading-response peak knee abduction moment, peak knee internal rotation moment, knee extension range of motion, and knee abduction range of motion, and increased loading-response and push-off response peak mediolateral GRF, push-off peak knee adduction moment, peak knee external rotation moment, peak knee abduction angle, and knee internal rotation range of motion. This study provides important information regarding differences in knee joint biomechanics during stair ascent between obese and healthy populations.

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# Chapter I

# Introduction

## Background

Obesity is a growing epidemic which involves an accumulation of an excessive amount of body fat, resulting in a body mass index (BMI) greater than 30kg/m$^2$ (31). Between 1960 and 1962, 13.3% of all adults (20-74 years) in the United States were classified as obese (21). In contrast, 34.9% of Americans were classified as obese in 2011-2012 (46). As of 2014, updated statistics suggest that 37.7% of American adults are now classified as obese (22). Moreover, the growing prevalence of obesity is not confined to the US. In 2014, the World Health Organization (WHO) indicated that there were 1.9 billion overweight and 600 million obese adults worldwide (72). Obesity has been associated with a range of negative health conditions, including increased risk for osteoarthritis, high blood pressure, dyslipidemia, heart disease, type 2 diabetes, respiratory dysfunction, and cancer (3, 5, 12, 52).

One major concern is the contribution of obesity to the incidence and progression of osteoarthritis (OA), especially in the knee. Previous research has found that obese participants were 6.8 times more likely to develop knee OA than healthy-weight participants (12). Other findings suggest that obese and overweight participants have increased odds ratios for developing OA (5) and for developing limitations on activities of daily living (ADL) (3). Two common ADLs are walking over level ground and using stairs. Several studies have investigated the biomechanics of obese participants during level-ground walking. These studies have shown that obese participants display larger peak external knee extension moment (6, 8), peak internal knee abduction moment (KAbM) (6), peak external hip extension moment (8), and peak external plantarflexion moment (20) compared, in absolute (non-normalized) terms, to healthy weight participants. Larger peak KAbM may impose greater loading to the medial knee compartment, which has been associated with the development and progression of medial compartment knee OA (5, 12).

In contrast to the research on level-ground walking, there is no empirical evidence about the lower-extremity biomechanics of obese participants engaged in stair ambulation. In healthy-weight participants, stair ascent has been found to significantly increase peak knee extension moments and peak external knee abduction moments compared to level walking (44). Another study with healthy-weight participants demonstrated significantly smaller first and second peak internal KAbMs during stair ascent compared to level walking (66).

Studies of level-ground walking have shown that obese participants have larger absolute (non-normalized) peak knee extension moments and KAbMs compared to healthy-weight participants (6). Stair ascent has also been shown to increase peak knee extension moments (44) and decrease peak KAbMs (66) in healthy weight participants. It is reasonable to expect that obese participants using stairs might show higher values for both variables, compared to healthy participants. It is currently unknown, however, how the knee biomechanics of obese participants differ from those of healthy-weight participants during stair ambulation.

Due to its association with increased medial knee loading and knee OA, several studies have examined how peak KAbM is affected by different gait modifications. One approach has investigated the effects of increased step width (SW). Wider SW has been shown to reduce peak knee extension moment and KAbM during level walking (4, 24, 76). A reduction in peak KAbM has also been demonstrated in stair negotiation in healthy (4, 49) and osteoarthritic (48) populations. In the stair ambulation studies (47-49), *Wide* and *Wider* SW conditions were set at 26% and 39% of the participant's leg length, respectively (38). A similar gait modification may be a helpful strategy in decreasing peak KAbM in obese participants during stair ambulation and ultimately reducing the risk of knee OA in this population.

**Statement of the Problem**

No studies have examined the differential effects of increased SW on knee biomechanics during stair ascent by obese and healthy-weight participants. Previous studies have shown significant differences between obese and healthy-weight subjects during level walking, while walking at preferred SW. It is unknown, however, if such differences will translate to stair ascent. Therefore, the purpose of this study was to compare knee biomechanics of obese and healthy-weight participants during stair ascent under preferred and increased SW conditions.

**Research Hypotheses**

1. The increased SW condition will display smaller peak KAbM regardless of weight group.
2. The obese group will display larger peak KAbM than the healthy-weight group.
3. The obese group will exhibit greater peak knee extension moments than the healthy-weight group.

**Delimitations**

The exclusion criteria included:

- Any major lower extremity injuries or surgeries.
- Any disorder/disease/pathology affecting gait or balance.
- Any lower extremity injuries within the past year.
- Any pain while performing common actives of daily living, such as walking or using the stairs.
- Any cardiovascular diseases or primary risk factor that prohibited participation in aerobic exercise as indicated by the Physical Activity Readiness Questionnaire (PAR-Q) If a

participant marked “yes” on any of the questions, he or she was required to provide written consent from a doctor signifying adequate health for participation in the study.

The inclusion criteria for obese participants included:

- Men and women between 18 and 40 years of age.
- BMI between 30kg/m$^2$ and 39.9kg/m$^2$.

The inclusion criteria for healthy-weight participants included:

- Men and women between 18 and 40 years of age.
- BMI between 18kg/m$^2$ and 24.9kg/m$^2$.

**Limitations**

- The tests were conducted in a laboratory setting.
- Skin marker placement in obese participants may not accurately reflect bony landmark location.
- The obese group was limited to a BMI of 39.9kg/m$^2$ because higher BMI levels decrease tracking accuracy of skin mounted markers.
- Reflective markers used to track the feet were placed on the shoe, and therefore might not have accurately reflected the motion of the foot within the shoe.

# Chapter II

# Literature Review

## Introduction

According to a recent study, 37.7% of American adults are classified as obese, as of 2014 (22). With increasing prevalence of the obesity epidemic, there is an increased need for research on the effects of the increased adiposity on the human body. This review will summarize the current literature on the effects of obesity on the kinematics and kinetics during level walking and stair ascent/descent, the effects of step width on lower extremity biomechanics, and adjustments to segment inertia parameters. The purpose of the proposed study from this literature review is to compare knee biomechanics of obese participants and healthy weight participants, and examine effects of increased step width on knee biomechanics of obese and healthy weight participants, during level walking, stair ascent, and stair descent.

## Level Walking: Obese versus Healthy Weight

The first major section of this review will discuss the impact of obesity on level walking. This will be divided into three separate sub-sections that will discuss the spatiotemporal, kinematic, and kinetic differences found in obese participants.

### *Spatiotemporal Characteristics*

Previous studies have shown obese participants to have different gait patterns than healthy weight participants (8, 18, 20, 23, 32, 58, 59, 62). Obese participants reportedly have a 0.3 m/s, or 16% (20), slower preferred walking velocity than healthy weight participants (18, 23, 32, 58, 60, 62). This reduction in walking speed is important because increases in walking speed increase vertical, anteroposterior, and mediolateral ground reaction forces (GRF) experienced in both healthy and obese participants (8). Additionally, increased walking speed increases peak hip flexion/extension moments, and peak knee extension/flexion moments (34).

This slower preferred walking velocity is likely due to a stride length that is an average of 0.26 m shorter (18, 58, 62) than healthy weight participants. When normalized to body height, obese participants had a significantly shorter, 0.06 m/m, stride length (32), this is significant because it means that obese participants would have shorter stride lengths than healthy weight participants of the same height. Obese participants also had a 0.11 m (18), or 7% (20), shorter step length compared to healthy weight participants. These findings are in conflict with the findings by Browning and Kram (8), who found no difference in stride length between obese and healthy weight participants at standardized speeds ranging from 0.5m/s to 1.75 m/s.

Unlike stride length, step length, and walking speed, obese participants have been shown to consistently increase step width. The magnitude of this difference is often contested. Spyropoulos et al. (62) found obese participants have up to two times greater step widths as healthy weight participants (0.16m vs 0.08m). However, Sarkar et al. (59) found obese male participants have a 42% greater step width and Browning and Kram (8) found obese participants had a 30% greater step width.

The final spatiotemporal characteristic is differences in walking cycle characteristics. At preferred walking speeds, obese participants spend significantly longer periods of time in the stance phase (+3.28%) and double support phase (+1.74%) (32, 62). At standardized speeds, obese participants also spent an average of 3.53% longer in the stance phase, and 6.27% longer in double support, during each speed, ranging from 0.50m/s to 1.75m/s (8).

To date, there have been many spatiotemporal differences found between healthy weight and obese participants. Obese participants have been reported to have: slower preferred walking velocities (18, 20, 23, 32, 58, 60, 62), increased GRFs (8), decreased stride lengths (18, 32, 58, 62), decreased step lengths (18, 20), increased step width (8, 59, 62), and increased time spent in stance and double support phases (8, 32, 62). The next section will investigate the kinematic differences found between healthy weight and obese participants.

*Joint Angular Kinematics*

Contrary to the spatiotemporal characteristics, which are more common among studies, many kinematic findings are in conflict with each other. For example, to date, most studies have reported no significant differences in knee extension/flexion angles between obese and healthy weight participants walking at preferred speeds (8, 23, 32, 62). However, DeVita and Hortobagyi (20) reported obese participants tend to walk with approximately 8˚ less knee flexion during early stance, and 4˚ less flexion throughout stance phase, each a significant decrease.

Differences between healthy weight and obese participants have also been found frontal plane knee motion. These differences are disputed, however, leading to an uncertainty of any true differences between healthy weight and obese participants. For example, Lai et al. (32) found obese participants have a 4.8˚ higher peak knee adduction angle during stance phase and a 10.0˚ higher peak adduction angle during swing phase. Additionally, the knee had an average abduction angle of 1.7˚ for healthy weight participants and 6.29˚ for obese participants throughout swing phase, an 8.0˚ difference. On the contrary, Silvernail et al. (23) found obese participants have a 3.8˚and 3.2˚ lower peak adduction angle while walking at preferred walking velocity and at 1 m/s, respectively, compared to healthy weight participants. The authors also reported that obese participants had an average peak knee adduction angle of 2.4˚ (at preferred speed) and 2.9˚ (at 1 m/s); each of these were significantly lower than the 6.2˚ (at preferred speed) and 6.1˚ (at 1 m/s) average peak knee adduction angles of the healthy weight participants.

To date, most research on the knee has been performed with the sagittal and frontal planes being the primary planes of interest. One study by Lai et al. (32) was unable to find any significant difference between obese and healthy weight participants in the transverse plane.

Previous studies have found that obese participants walk with different hip joint kinematics than their healthy weight counterparts (20, 32, 62). At a standardized speed of 1.5 m/s, DeVita and Hortobagyi (20) found that obese participants' hips were approximately 5˚ more extended throughout stance phase, causing amore erect posture than healthy weight individuals (20). However, Browning and Kram found no difference in hip angle during midstance between obese and healthy participants at any standardized speed (8).

While walking at self-selected speeds, Spyropoulos et al. (62) and Lai et al. (32), found no significant differences in hip flexion/extension angles throughout stance phase. However, they did find that obese participants had increased average hip abduction angles at midstance (+9.5˚), opposite heel strike (+20.2˚), and toe off (+13.0˚) (62), and increased hip adduction angles at terminal stance (+3.2˚) and pre-swing (+3.7˚) (32). Due to the difficulty of assessment in obese participants, the hip is not often the subject of intensive research.

In addition to the hip and knee, the ankle is also commonly researched in studies on lower extremity kinematics. The differences in the kinematic variables of obese and healthy weight participants associated with the ankle are also commonly disputed. Lai et al. (32) and Browning and Kram (8) found no significant differences in peak dorsiflexion angles while walking at preferred walking velocity and at standardized velocities between 0.5 m/s and 1.75 m/s, respectively. Differences were found, however, by DeVita and Hortobagyi (20), who reported that obese participants were approximately 6˚ more plantarflexed throughout stance phase compared to healthy weight participants.

In the frontal plane, Lai et al. (32) found obese participants have ankle kinematics that are significantly different than those of healthy weight participants. Most researchers chose to find the peak angle and ROM throughout stance, which is typically considered the best way to describe ankle motion. Lai et al. (32) found peak angles at three different points of the stance phase, which may be problematic as these points do not necessarily coincide with the peak angle. At mid-stance, healthy weight participants demonstrated 2.1˚ of inversion, while obese participants demonstrated 2.2˚ of eversion. At terminal stance, each group of participants became more everted, however, healthy weight participants were at 0.85˚ of inversion, and the obese participants were at 4.3˚ of eversion. At pre-swing, participants became more inverted, healthy weight participants had 3.0˚ of inversion and obese participants were everted by 1.6˚. Additionally, Messier et al. (42) studied the impacts of obesity on the foot, and demonstrated that obese participants make initial foot contact with greater ankle inversion (+5.6˚ of inversion), have a greater maximum eversion angle (+1.0˚ of eversion), and, therefore, have a 6.6˚ greater range of motion than healthy weight participants.

The ankle is the only lower extremity joint that has significant research performed involving the transverse plane. The motion of the foot in the transverse plane is commonly referred to as toe-in and toe-out. Sarkar et al. (59) studied the effects of obesity on balance and gait, and how it effects males and females differently and found that obese participants tended to walk with greater toe-out angle than healthy weight participants (+1.5˚ for females, and +3.2˚ for males). This finding was supported by Messier et al. (42), who demonstrated that obese participants walked with 4.4˚ greater toe-out angle compared to their healthy weight counterparts.

To date, there are many kinematic differences between obese and healthy weight participants. Many of the differences discussed here continue to be debated, as conflicting results continue to arise from research. Overall, research has shown that obese participants may have more extended hips through stance phase (20), increased hip abduction from midstance to pre-swing (32, 62), less knee flexion throughout stance (20), more abducted knee during stance (23, 32), more plantarflexion throughout stance and at toe-off (20), more inverted ankle at contact (42), more everted foot from midstance to pre-swing (32), greater peak foot eversion (42), and an increased toe-out angle while walking (42, 59). The next section will cover the changes in the kinetics of healthy weight and obese participants.

*Joint Kinetics*

For the purpose of this literature review, discussion on joint kinetics will be focused on the moments experienced at a joint. A joint moment is caused by muscular efforts to generate and control movements at the joints and GRF vector passing perpendicular to the axis of rotation. Joint moments can be evaluated using either internal or external techniques. An external joint moment is the moment generated through exterior forces, such as GRFs. An internal moment is generated by the mechanisms within the body (i.e. muscles) producing a torque to generate movement.

The method of evaluating joint moments is disputed among researchers studying the obese population. This is because obese participants are expected to have higher GRFs due to their higher body weight, and therefore have increased joint moments. Since we know these values are different, normalization is used to eliminate the effects of body mass on joint moments. Typically, researchers may either normalize joint moments by body mass or by lean body mass. Normalization by lean body mass, similar to normalization by total body mass, will eliminate the effects of lean body mass on joint moments, and therefore allows the researcher examine the effects of increased adiposity on the subject's joint moments, while normalization by body mass does not distinguish between lean and fatty tissues.

The substantial differences between normalizing and not normalizing can be found at each of the three primary lower extremity joints, hip, knee, and ankle. These differences in methods lead to a variety of conflicting results. At the hip, Browning and Kram (8) found, prior to normalization by body mass, obese participants had increased hip extensor moment by an average of 42.9 Nm greater at each speed. However, after normalization, this difference disappeared. This is similar to the findings of DeVita and Hortobagyi (20) and Lai et al. (32) who both found that there were no differences in peak sagittal (20, 32), frontal (32), and transverse (32) plane hip moments between obese and healthy weight participants after normalization.

The knee, one of the most commonly researched joints in biomechanics, is also commonly researched in the obese population. This is because obesity has been found to be a risk factor for knee osteoarthritis (33, 41). Unfortunately, the magnitudes of the kinetic variables of the knee are also disputed, largely due to inconsistency in methods of normalization, or a lack of normalization.

Prior to normalization, Browning and Kram (8) found obese participants to have significantly higher knee extensor moments at 1.75 m/s, and 51% (non-significant) greater peak knee extensor moments at 1.50 m/s. However, these differences disappeared following normalization by body mass This is similar to the findings by Silvernail et al. (23), who found no differences in the peak external knee flexion moment once normalized to lean body mass and height. Lai et al. (32) also found no differences in peak knee flexion moment between the groups when normalized to body mass and height. While these two different types of normalizations were able to come to the same conclusion, DeVita and Hortobagyi (20) found that when normalized to body mas, obese participants had a significantly lower peak knee extensor moment than healthy weight participants while walking at the same speed. This study also demonstrated that BMI has a strong, inverse ($r = -0.70$) relationship with the peak sagittal plane knee moment during the stance phase of gait. Starting at approximately 30 $kg/m^2$, the peak knee moment during stance becomes increasingly flexor as BMI continues to increase. This decrease in extensor moment may be linked to the differences in the spatiotemporal characteristics previously discussed. DeVita and Hortobagyi (20) suggest that obese participants may alter their gait, by increasing their peak ankle plantarflexion moment and power, and decreasing their peak knee extensor moment over the stance phase, to reduce knee joint loading.

Frontal plane knee kinetics are also often studied due to the relationship with knee osteoarthritis (23, 35, 60). A common variable is the external knee adduction moment (EKAdM) which is equivalent to the KAbM. Increases in KAbM have been associated with increased risk of knee osteoarthritis (35). Blazek et al. (6) found that obese participants had higher absolute peak knee flexor and adduction moments. However, after normalization, the flexor moment showed no differences between groups, and the adduction moments became significantly lower than those in the healthy weight group. This is in contrary to the findings by Silvernail et al.(23), Lai et al. (32), and Segal et al. (60), who found no differences of the variable between groups after normalization.

Blazek et al. (6), who studied obese participants at various ages, found that normalized peak EKAdM of obese participants significantly increases (small increase, but significantly) with age ($R^2$=0.19, p=0.007). This is interesting because it shows that as obese participants age, they become unable to decrease their adduction moment via gait alterations, and the peak EKAdM, which was once significantly lower than healthy weight participants, becomes similar with that of healthy weight participants of the same age.

Segal et al. (60) studied different forms of obesity, and compared them to healthy weight participants. This was done in order to determine if the location of increased adiposity significantly altered the EKAdM. They found that an increased mass was responsible for changes in peak EKAdM. No differences were found, however, between the two obesity types, android and gynoid. Android obesity occurs when participants tend to keep their excess fat mass in the chest and torso areas, giving them an apple shaped appearance. Gynoid obesity, however, occurs when participants hold most of their excess fat mass in the hips and legs, giving them a pear shaped appearance.

To the author's knowledge, little research has been done in the transverse plane of the knee. Lai et al. (32) states that they did not find any differences in peak knee internal/external rotation moments. The lack of research in this area may be because it has not been associated with joint problems, such as knee osteoarthritis, or because it is difficult to observe differences in this plane due to minimal movements.

For ankle kinetics, although it was studied less than the knee, significant differences have been found in the sagittal and transverse planes of the ankle. DeVita and Hortobagyi (20) demonstrated that obese participants had a significantly increased peak plantarflexor moment compared to healthy weight participants, prior to normalization. After normalization by body mass and height, Lai et al. (32) showed that obese participants had a significantly decreased peak ankle plantarflexion moment (-0.12 Nm/kg/m), and a significantly increased peak inversion moment (+0.04 Nm/kg/m). This is similar to what Browning and Kram (8) found who showed that peak ankle plantarflexor moments were significantly lower in obese participants than in healthy weight participants at all walking speeds.

To date, there are many differences that have been found between obese and healthy weight participants' kinetics while walking on level ground. As stated previously, many of these differences are highly debated. The differences can most easily be broken down into two types: non-normalized and normalized.

Non-normalized differences currently found in the literature are: increased peak hip extensor moment (8), increased peak knee extensor moment (6, 8), increased peak knee adduction moment (6), and an increased peak plantarflexion moment (20) in obese participants. When data is normalized by body mass or body mass*height, obese participants have a: decreased peak knee extension moment (20), decreased peak knee external adduction moment (6), decreased peak ankle plantarflexion moment (8, 32), and an increased peak inversion moment. Normalization by body mass and height did not influence the significant differences compared to normalization via body mass only. Additionally, it has been found that the location of increased fat-mass did not affect the peak knee external adduction moment. The additional weight of the subject (60), however, is the variable that most heavily influences the peak EKAdM. Additionally, this moment increases significantly with age of obese participants, but not of healthy weight (6).

**Level Walking: Effects of Added Mass**

In order to provide a more comprehensive view of the effects of increased mass on the body a few studies have added mass onto healthy weight participants to simulate obesity. These types of studies typically provided a sudden mass gain via mass added to foot (9), shank (9), thigh (9, 71), and waist (9). Additionally, Westlake et al. (71) researched the effects of added thigh circumference on knee biomechanics in addition to the added mass. To date, very few studies have studied effects of added mass on gait.

*Spatiotemporal Characteristics*

When walking with a 4kg load placed on the foot, Browning et al. (9) found that participants had a significantly greater stride length (+0.08m), a significantly slower stride rate (-9%, while walking at 1.25m/s), and a significantly slower swing time (+0.04s), compared to all other loaded conditions (no load, waist load, thigh load, and shank load). When an additional 4kg was added to the foot, they found stride length significantly increased a further 0.08m, stride rate decreased 0.04Hz (while walking at 1.25m/s), stance time became significantly greater (0.05s or 10%), and swing time increased significantly by an additional 0.03s compared to the 4kg foot load condition.

Westlake et al. (71) studied the effects of added mass, added circumference, and both added mass and added circumference (combination) to the thighs while participants walked at their preferred walking speed (average: 1.29±0.15m/s). They reported the step width outcomes for female and male participants separately. Both male and female participants had an average width of 0.08m during the control trials. For females, the added circumference and combination conditions both lead to a statistically significant increase in step width, with an average of 0.11m in each condition. These two conditions were larger than that of the added mass condition, which increased by 0.01m, which was still significant. For males, increased circumference had the strongest effect on step width, increasing the average width up to 0.13m. The combination condition had the second highest effect in the male participants, leading to a 0.11m average step width. In the male group, there was no difference in step width between control and mass only trials, with both being 0.08m.

*Joint Angular Kinematics*

Due to a lack of literature on this subject, the availability of kinematic data is limited. To the author's knowledge, only one study exists that attempts to look at the kinematic differences. Westlake et al. (71), who studied the effects of added mass and circumference on the thighs, found no significant differences in the peak knee flexion angle between the control condition and the added mass, added circumference, and combination conditions. However, there was a significant difference between added circumference and added mass and combination conditions; the added circumference alone increased the peak knee adduction angle significantly higher than the added mass and combination conditions. Due to a lack of additional studies on the kinematic changes when mass is added to the lower extremities, further research is needed.

*Joint Kinetics*

Kinetically, Browning et al. performed research on the effects of mass added to the waist, thigh, shank, and foot during level treadmill walking at 1.25m/s, and found that the only significant changes occurred when mass was added to the foot. At the hip, Browning et al. (9) found that when 8kg was added to the foot, participants had significantly greater peak hip extensor (+0.32Nm/kg) and peak hip flexor (+0.42Nm/kg) moments, these differences disappeared during all other loading conditions.

Browning et al. (9) also found that the peak knee flexor moment was significantly increased by 54% with 4kg added mass, and 100% with 8kg added mass. They found no additional kinetic differences for any other loading conditions. This is in agreement to Westlake et al. (71) who found that an increase in thigh circumference (males) and an increase in circumference, mass, and a combination of the two (females) significantly increased the peak knee extension moments over the control condition. However, in males, the increased mass and combination conditions resulted in a lower peak knee extension moment (significance not reported). Additionally, they found that peak KAbM were similar across all conditions.

This review found that no kinematic differences have been when mass and/or circumference is added to a participants lower extremities; however, kinetic differences have been found with increased hip flexor and extensor moments (9) and increased knee flexor moments (9), and there are kinetic differences between added mass and added circumference.

Spatiotemporally, when 4kg is added to a subject's foot, they tend to have an increased stride length (9), decreased stride rate (9), and slower swing time (9). When an addition al 4kg is added, these changes become larger, and stance time also becomes significantly longer (9). It has also been found that added mass, added circumference, and a combination of the two onto the thigh, will significantly increase step width (71). Additional research on the effects of added mass or circumference on lower extremities are warranted to gain a better understanding of these effects on lower extremity biomechanics.

**Level Walking: Effects of Weight Loss**

In the previous section, we examined the effects of added mass and circumference on participants during gait. This section will focus on the opposite. This section will ask the question, "What happens if a subject undergoes significant weight loss?" The participants involved in this section were obese and then underwent significant weight loss induced by surgery (30, 70). Participants were then either compared to healthy weight participants (30) or to themselves pre- and post-surgery (70). This is important because this will help broaden our knowledge of how obesity affects the body by researching not only what occurs when participants gain significant fat mass, but also when a significant amount of fat is lost.

*Spatiotemporal Characteristics*

The effects of weight loss on the spatiotemporal characteristics in gait are relatively unknown. Since gaining fat mass has been shown to change gait characteristics, it is possible that loss of this fat mass may lead to additional changes in a subject's gait, and these change may or may not cause their gait characteristics to become more similar to healthy weight participants.

Hortobagyi et al. (30) studied that effects of massive weight loss, due to metabolic surgery, on participants instructed to walk at a self-selected speed and at 1.5m/s on a level surface. Gait kinematics and kinetics were recorded before, at 7.0 (±0.7) months, and at 12.8 (±0.9) months after surgery. They found that when participants underwent massive weight loss, preferred walking speed increased. The participants increased their walking speed by 3.9% after the first 27% of weight was lost. When an additional 6.5% of weight was lost, there was a 7.3% increase in walking velocity. This is expected because obese participants have been found to have slower preferred walking speeds compared healthy weight participants (18, 20, 32, 58, 60, 62).

Hortobagyi et al. (30) also found that the participants that underwent massive weight loss increased their stride length while walking at a standardized speed of 1.5m/s. This is in agreement with previously discussed research that found that when compared to healthy weight participants, obese participants had a shorter stride length (18, 32, 58, 62). However, Vartiainen et al. (70) did not find any significant differences in stride length when comparing participants pre and post bariatric surgery at 1.2m/s and 1.5m/s.

Another difference found within this type of characteristic is in the step width of the participants. Vartiainen et al. (70) found that post bariatric surgery participants significantly decreased step width by an average of 0.035m. This difference is important because previous studies have found that obese participants walk with a significantly, 0.1m (18, 62) or 42% (59), greater step width than healthy weight participants.

Lastly, Vartiainen et al. (70) and Hortobagyi et al. (30) studied the gait cycle parameters of the participants who underwent massive weight loss. Hortobagyi et al. (30) found that participants who underwent weight loss spent significantly less time in swing phase than healthy weight participants. No other differences in gait cycle characteristics were found in this study. Vartiainen et al. (70), however, found no significant differences in any cycle characteristics from pre and post bariatric surgery induced weight loss. This is unexpected, as we have previously discussed, obese participants spend significantly longer time in stance (32, 62) and double support time (32) than healthy weight participants. We would therefore expect participants who have lost significant fat mass to have decreased time spent in stance and double support.

While there have been few studies that attempted to identify the effects of massive weight loss on the spatiotemporal characteristics of participants' gait, the current literature suggests that there are a few differences. This review found that when participants lose massive amounts of weight, their walking velocity increases (30) and their step width decreases (30). Other changes, such as increased stride length and changes to cycle characteristics are disputed in the literature (30, 70).

*Joint Angular Kinematics*

To date, there is little information on how weight loss affects walking gait kinematics. The information that is available (30, 70) has inconsistent results. In the hip, Hortobagyi et al. (30) found that at a self-selected speed, weight loss produced a significantly increased sagittal plane hip ROM (difference between most extended and most flexed positions of the hip during gait). This significant difference disappears when the participants walk at the standardized speed, 1.5m/s. This increase in ROM is in contrast to Vartiainen et al. (70) who demonstrated that participants had a significantly decreased hip flexion angle at initial foot contact, and therefore less ROM.

Hortobagyi et al. (30) also examined the sagittal plane kinematics of the knee. The study demonstrated that massive weight loss increased maximal knee flexion in early stance while walking at a preferred walking velocity. This change again disappeared when walking at a standardized speed. This change is disputed by Vartiainen et al. (70) who found no difference in maximal knee flexion at early stance. Hortobagyi et al. (30) also found that these participants had an average angular position that was significantly less plantarflexed during both preferred and standardized walking speeds.

To the author's knowledge, there are currently only two studies that have researched the kinematics of participants who have underwent massive weight loss. It is demonstrated that these participants may have increased hip ROM, maximal knee flexion, and more plantarflexed average angular position of the ankle (30); however, these findings are disputed by the only other study (70). Additional research is needed to better understand the effects of massive weight loss on gait kinematics.

*Joint Kinetics*

The two studies that were involved in studying the kinematic changes caused by weight loss also studied how this weight loss affected the sagittal and frontal plane joint moments (30, 70). Vartiainen et al. (70) found significant differences were found between pre-surgery and post-surgery in both the hip and knee. At the hip, participants had significantly, 30Nm, lower peak hip extensor moment at 1.2m/s and a 39Nm reduced peak hip extensor moment at 1.5m/s. These differences, however, disappeared after normalization by body mass and height (70).

Hortobagyi et al. (30) found no difference in the peak hip moment in the sagittal plane when comparing pre-surgery to 7 and 12 months post-surgery. Also in the sagittal plane, Vartiainen et al. (70) found a 11Nm decrease in the peak knee flexor moment while walking at 1.2m/s and a 18Nm decrease while walking at 1.5m/s, prior to normalization. No differences were found after normalization. Hortobagyi et al. (30), however, found that after normalization by body mass and height, the peak knee extensor moment in early stance significantly increased, by 1.0N/kgm while walking at self-selected speed from pre-surgery to the seven month follow-up. Additionally, while walking at self-selected speed, the peak knee extensor moment significantly increased from the 7-month follow-up to the 12-month follow-up.

In the frontal plane, Vartiainen et al. (70) determined that the peak EKAdM during early stance and late stance was significantly decreased, 13Nm and 10Nm respectively, while walking at 1.2m/s. Additionally, the study found that while walking at 1.5m/s, the peak EKAdM during early stance significantly decreased by 9Nm prior to normalization. The differences at 1.2 m/s disappeared after normalization; however, the reduction during early stance, while walking at 1.5m/s, still significantly decreased by 0.04Nm/kgm after normalization. Hortobagyi et al. (30) also found that the KAbM was significantly decreased at both standardized (1.5m/s) and self-selected speed; however, these differences disappeared after normalization. Hortobagyi et al. (30) also found that the peak dorsiflexion moment was significantly decreased after weight loss at both speeds, and the differences disappeared following normalization.

To date, there have been only two studies that have reported on changes in a subject's gait due to massive weight loss. These two studies are often in conflict with each other in differences reported. These studies do suggest that massive weight loss will: increase walking velocity and decrease step width (30), reduce absolute peak hip extensor moments (70), decrease in peak knee flexor moment (70), and decrease peak EKAdM during early and late stance (70). Further research is needed to strengthen our knowledge of the effects of massive weight loss on gait parameters. This will also help us understand how obesity affects participants while walking.

**Stair Negotiation**

The effects of stair negotiation on healthy/young individuals on lower extremity biomechanics has been researched for years due to being a common activity of daily living for adults old and young (67). A recent review by Standifird et al. (64) suggests that many studies researching lower extremity joint mechanics on stairs tend to have different number of total steps in their staircases, and different steps of interest (SOI). For example, one of the first studies to research lower extremity mechanics while ascending and descending stairs was performed by Andriacchi et al. (1) in 1980. This study was performed on a three-step stair case with GRF instrumented on the first-step, or a first-step SOI. Later, McFadyen and Winter (39) used a five-step staircase with the second-step as the SOI. Later research on stair gait varies widely on the number of instrumented stairs used for research, two studies by Mandeville et al. (36, 37) used only one instrumented step, while a study by Wilken et al. (73) used sixteen instrumented steps. Additionally, the step(s) of interest vary greatly from the first-step as SOI (1, 13, 36, 37), second-step as SOI (39, 45, 54), third-step as SOI (48), fourth-step as SOI (26), steps one and two as SOI (44), steps one through three as SOI (49, 50, 57, 61, 65, 74), steps one and three (ipsilateral limb) as SOI (69), steps three and four as SOI (68), and steps five through eight as SOI (73).

Work by Yu et al. (74) focused on the reproducibility of stair gait kinetics and kinematics. This study found that the greatest variability in lower extremity joint angles and moments occurred during the first-step of ascent on a staircase. This first-step was deemed as a transitional step between level walking and the initiation of stair ascent (74). However, the actual magnitude of the differences in biomechanical variables between the first-step and subsequent steps was not reported.

In order to understand the magnitude of the changes in lower extremity biomechanics between steps during stair ascent, Standifird and Zhang (66) recently researched the first three steps of an instrumented staircase, and level walking in healthy, middle aged to older (45-68 adults. This study found that, in the sagittal plane, the ankle had significantly greater dorsiflexion at contact (18.1±7.1º) on step two than on step one (10.7±7.2º), plantarflexion ROM on step two (-35.5±8.4º) was greater than that on step one (-26.0±8.5º), and first peak plantarflexion moment on step two (-0.7±0.3Nm/kg) and three (-0.8±0.3Nm/kg) were greater than on step one (-0.6±0.3Nm/kg). No statistically significant differences were found between step two and step three at the ankle in the sagittal plane (66).

In addition, it was found that the knee was significantly more flexed at contact of steps two (-69.1±4.6º) and three (-70.7±5.9º) compared to step one (-57.3±5.6º), extension ROM was significantly greater for steps two (57.3±4.0º) and three (59.3±3.6º) than step one (44.5±7.1º), and the second peak knee extension moment at step three (0.4±0.3Nm/kg) was significantly lower than step one (0.5±0.3Nm/kg). In the sagittal plane, no significant differences were found between steps two and three at the knee. However, the peak flexion moment at the hip was significantly lower during step three (0.4±0.2Nm/kg) than step two (0.5±0.2Nm/kg) and step one (0.5±0.2Nm/kg); steps one and two of this variable were not statistically different (66).

In the frontal plane, the ankle had a significantly increased first peak inversion moment during step one (0.3±0.1Nm/kg) compared to step two (0.3±0.1Nm/kg), while the knee experienced a significant reduction in the first peak KAbM from step one (-0.4±0.1Nm/kg) to step two (-0.3±0.1Nm/kg) (66). Additionally, the hip experienced a significant reduction in the first peak abduction moment from first (-0.9±0.2Nm/kg) to second (-0.7±0.2Nm/kg) step, and a reduction in the second peak hip abduction moment from first (-0.6±0.2Nm/kg) to second (-0.5±0.1Nm/kg) steps (66).

Unlike the sagittal plane in this study, there were many frontal plane variables that differed significantly between step two and step three (66). At the ankle, step two had a significantly higher: first peak inversion moment (0.3±0.1Nm/kg) than step three (0.3±0.1Nm/kg), and second peak inversion moment (0.4±0.1Nm/kg) than step three (0.2±0.1Nm/kg). At the knee, step two had a significantly lower first peak KAbM (-0.3±0.1Nm/kg) than step three (-0.4±0.1Nm/kg). At the hip, the first and second peak abduction moments for step two (-0.7±0.2Nm/kg and -0.5±0.2Nm/kg, respectively) were both significantly lower than those for step three (-0.9±0.2Nm/kg and -0.6±0.2Nm/kg, respectively).

These two studies (66, 74) demonstrate the importance of the SOI on the interpretation of results of stair gait data. In addition to these findings, Vallabhajosula et al. (69) found that there are also biomechanical differences between step one and step three (second ipsilateral step), and if a subject approaches the stair from a walk versus a stand. This study found that initiating stair ascent from a walk lead to an increase in the peak KAbM on step one (0.8±0.1Nm/kg) compared to that from standing (0.7±0.1Nm/kg). There was also significant increases in the peak hip abductor moment on step one from initiating from a stand (0.9±0.1Nm/kg) compared to from a walk (1.0±0.1Nm/kg) (69). Similar to the study by Standifird and Zhang (66), this study found significant differences between step one and subsequent instrumented step.

They reported that the second ipsilateral step had a significantly higher peak ankle abductor moment (0.3±0.0Nm/kg) than the first (0.2±0.0Nm/kg). Additionally, the peak internal knee abductor moment during step two (1.0±0.1Nm/kg) has significantly higher than during step one (0.8±0.1Nm/kg). Furthermore, an increase in the peak hip abductor moment was found during step two (1.2±0.1Nm/kg) compared to step one (1.0±0.1Nm/kg) (69).

Due to the findings of these studies, it is clear that SOI needs to be established prior to data collection. Step one was found to have the most variability from subsequent steps (74), there are biomechanical differences between the first three steps of stair ascent (66), and there are significant kinetic differences when initiating stair ascent from a walk versus a stand (69).

*Stair Negotiation: Effects of Age*

Much of the current literature on stair gait involves participants at different ages. For example, Standifird et al. (64-66) and Paquette et al. (48-50) studied the lower extremity biomechanics of older populations, Costigan et al. (13) and Protopapadaki et al. (54) had young adult populations, and Strutzenberger et al. (68) studied children. Therefore, it may be important to study the effects of age on lower extremity biomechanics. Novak and Brouwer (45) examined differences between young (23.7 ± 3.0 years) and older (67.0±8.2 years) adults during both stair ascent and descent. Younger adults ascended the stairs with a significantly faster cadence (102.5±8.9 steps/min) than older adults (94.8±13 steps/min). Young adults also descended the stairs with a significantly higher cadence (110.6±10.2 steps/min) than older adults (103.7±15.6 steps/min). All kinetic variables associated with this research were reported as internal moment.

At the knee, young adults had a significantly higher peak flexion moment (0.3±0.1Nm/kg) than older adults (0.2±0.1Nm/kg) during ascent (45). During descent, younger participants had a lower second peak knee extension moment (1.0±0.2Nm/kg) compared to older adults (1.2±0.2Nm/kg). No significant differences between groups occurred in the frontal plane at the knee.

At the ankle, the young adults had a significantly higher peak plantarflexion moment (1.3±0.2Nm/kg) compared to the older adults (1.2±0.1Nm/kg), as well as higher first peak (0.9±0.2Nm/kg) and second peak (1.3±0.2Nm/kg) plantarflexion moments compared to the older adults (0.8±0.2Nm/kg and 0.7±0.2Nm/kg, respectively), during stair ascent (45). During stair descent, the young adults had significantly higher first peak plantarflexion moment (1.0±0.2Nm/kg) than older adults (0.8±0.2Nm/kg). No significant differences were found in frontal plane kinetics at the ankle.

The young participants also had a lower peak hip flexor moment (0.1±0.1Nm/kg) compared to older adults (0.2±0.1Nm/kg) during stair ascent (45). The young participants, however, had a higher peak hip flexor moment (0.4±0.1Nm/kg) than older adults (0.3±0.1Nm/kg) during stair descent. In the frontal plane, young participants had a lower second peak hip abduction moment during both stair ascent and descent (0.4±1Nm/kg and 0.5±0.1Nm/kg, respectively) compared to older adults (0.5±0.1Nm/kg and 0.6±0.2Nm/kg, respectively).

This study was one of the first studies to show significant lower extremity kinetic differences between young and old adults during stair gait. Due to the differences found, it can be expected that the results of a study on healthy and obese young adults should most closely resemble those of studies with similar age groups.

*Stair Negotiation: Effects of Step Width*

A common spatiotemporal measurement used in biomechanical research is step width. This variable was discussed often in previous sections of this literature review. However, the effects of step width on lower extremity biomechanics was not discussed. This variable is often modified in studies of gait biomechanics with knee osteoarthritis patients (24, 48-50, 56) because the KAbM, a surrogate measure of loading to the medial knee compartment, has been shown to increase with severity of medial knee osteoarthritis (43). Current literature examines effects of increased step width on lower extremity biomechanics in level walking (76), running (2, 7, 53), stair ascent (48), and stair descent (49, 50).

Zhao et al. (76) reported results in an in vivo study with an instrumented total knee replacement implant of a single subject during level walking. This study showed a decrease in the EKAdM from preferred width (2.6±0.2% BW x HT) to wide step width (2.4±0.2% BW x HT) (BW=body weight, HT=Height). Additionally, this study found that there was an increase in medial knee contact force from preferred step width (1.6±0.1BW) to wide step width (1.7±0.1BW). However, the size of step width increase was not reported.

While running on level ground, Brindle et al. (7) found that increasing step width significantly lowers peak hip adduction angle, decreases peak rearfoot eversion angle, decreases peak KAbM, and decreases peak rearfoot inversion moment. Arellano and Kram (2) found that as subject increased their step width, there was an apparent positive linear relationship between step width and metabolic demand, as step width increased, the metabolic demand also increased.

Studies focusing on the changes in lower extremity biomechanics in stair ambulation gait due to increases in step width have so far been limited to changes in the knee. To date all studies on this subject have been performed by Paquette et al. (48-50), who researched step width effects on healthy older adults (58.9±8.3 (48), 54.8±8.9 (49)) and older adults with knee osteoarthritis (62.5±9.0 (48, 50)). For the purpose of this literature review, only results of the healthy participants will be discussed.

During stair ascent, Paquette et al. (48) studied how an increase in step width to 26% (wide) and 39% (wider) of leg length affected the KAbM. Healthy participants had a preferred step width of 0.1±0.0m, a wide step width of 0.2±0.0m, and a step width of 0.3±0.0m in the wider condition. They found that the first peak KAbM was significantly reduced in the wide step width (-0.3±0.8Nm/kg) from the preferred step width (-0.4±0.01Nm/kg). The second peak KAbM was also significantly reduced with increased step width: preferred (-0.3±0.1Nm/kg) was significantly higher than wide (-0.2±0.1Nm/kg)(48). The first and second peak KAbM was also significantly reduced in the wider condition, compared to preferred, but these values were not different than the wide condition.

Stair descent with increased step width also showed significant changes in knee kinematics and kinetics. Paquette et al. (49) found that there was a significant decrease in the first and second peak adduction angle from preferred (5.9±2.6º and 8.4±4.5º, respectively) to wide (26% leg length) (4.7±2.9º and 6.0±2.6º, respectively) step widths. This significant decrease decreased further with a wider (39% leg length) step width which resulted in a first peak adduction angle of 4.6±2.8º and a second peak adduction angle of 4.9±2.8º. Participants in this study had a preferred step width of 0.2±0.0m, a wide step width of 0.2±0.0m, and a wider step width of 0.3±0.0m, each statistically different. The first peak KAbM was, again, significantly reduced during the wide step width (-0.7±0.2Nm/kg) compared to preferred step width (-0.8±0.2Nm/kg). The second peak KAbM was found to be significantly lower during both wide (-0.4±0.1Nm/kg) and wider (-0.4±0.1Nm/kg) step width compared to preferred step width (-0.5Nm/kg±0.1Nm/kg), the second peak knee adduction angle and the second peak KAbM during the wide condition were significantly different from the wider condition.

The current literature on the effects of step width modifications suggest that an increase in step width while using stairs (48, 49) may cause a decrease in the first and second peak KAbM and decrease the first and second peak knee adduction angle. However, as previously discussed, there are known differences in the lower extremity biomechanics of healthy young and healthy older adults when using stairs. Furthermore, different results may be expected between young obese participants and their healthy counterparts when ascending and descending stairs with increased step widths.

To our knowledge, no studies have reported differences in step widths of young adult obese participants, who are otherwise healthy, compared to healthy-weight young adults. However, previous studies have reported that during level walking, obese participants consistently walk at increased step widths (8, 59, 62). The preferred steps widths of obese participants range from 30% greater (8) (exact values not reported) to 100% greater (62) (0.16m compared to 0.8m) than preferred step widths of healthy weight participants while walking on level ground. Sarkar et al. (59) found that obese males have a significant increase (42.2%) in preferred step width, but females had a non-significant decrease (13.4%) in preferred step width compared to healthy weight participants. Strutzenberger et al. (68) did not find any changes in step width in children during stair gait; however, this does not mean that obese young adults will have a similar step width as healthy weight young adults during stair gait.

Due to a lack of literature regarding obese participants while negotiating stairs, it is uncertain how the use of stairs will affect the preferred step width of obese participants during stair ambulation. It is likely that these participants who have a wider step width during level walking will continue to have a wider preferred step width while using stairs, however, this has not yet been tested. In order to test effects of increases in step width in obese participants, preferred step widths must be determined by taking data (subject height and step width) from previous work (8, 48, 49). Therefore, this study will express step widths as a percentage of a subject’s total height.

It was found that obese participants tend to walk with a SW that is between 7.4% and 8.0% (average: 7.7%) of their body height (8), while walking at 1.0m/s and 1.5m/s, respectively. This is greater than healthy weight participants who tend to walk with an average SW between 6.3% and 6.9% (average: 6.6%) (8), while walking at 1.0m/s and 1.5m/s, respectively. However, healthy weight participants tended to walk at a step width of 7.7% of their body height in stair ascent, and at 10% in descent at their preferred walking speed (average: 0.6m/s ascent and descent) (48, 49).

*Stair Negotiation: Effects on Healthy Weight Participants*

As previously discussed, the purpose of this literature review was to examine the effects of obesity and step width on a subject during level walking and stair gait. Specifically, this review is in preparation for research on the effects of obesity and step width on lower extremity joint kinematics and kinetics during level walking and stair gait. Now that the effects of obesity on level walking biomechanics has been carefully reviewed, we can examine the second area of this review: stair negotiation. In order to understand how obesity affects participants during this task, it is essential to understand how it affects healthy weight participants. Therefore, this section will cover the spatiotemporal, kinematic, and kinetic values of healthy weight participants while ascending and descending stairs, and how these values are different than level walking.

Due to the fixed size of a stair case, most studies have focused primarily on the speed of stair ascent/descent and cycle characteristics. Protopapadaki et al. (54), reported that healthy/young adults (28.1±6.1 years) ascend stairs significantly slower than descend. They found that participants would ascent at an average of 0.5m/s, while they would descend at 0.6m/s (±0.1m/s). This speed of ascent found by Protopapadaki et al. (54) is similar to the ascent speeds found by Nadeau et al. (44), whose participants were middle aged to older (41-70 years) adults, and ascended at 0.5m/s (±0.1m/s). The average speed of descent found by Protopapadaki et al. (0.6m/s), is in agreement with Paquette et al. (49) who determined an average descent speed of 0.6m/s (±0.1m/s), in healthy/older adults (54.8±8.9 years). These values are much lower than those reported by Standifird et al. (65), who reported an average ascent speed of 0.8 (±0.2) m/s in healthy, older/middle aged (62.3±7.5 years) individuals, which is higher than the values reported in healthy adults of similar age (58.9±8.3 years) by Paquette et al. (48) who ascended at 0.60±0.06m/s. The difference in average velocity can also be seen in cycle duration. Protopapadaki et al. (54) found that participants had a cycle duration that was 0.2 seconds longer (1.5s compared to 1.3s) when ascending stairs, which was a significant difference.

Nadeau et al. (44), compared spatiotemporal variables between level walking and stair climbing, in adults over 40 years old, in order to find the differences in these two modes of walking. They found that participants walked at a significantly reduced speed and cadence while ascending stairs (0.46m/s and 93.6steps/min) compared to level walking (1.16m/s and 105.4steps/min). Additionally, this study found that participants had (ascent versus level walking): a significantly longer total cycle time during stair ascent (1.3s versus 1.2s), significantly shorter stride length (0.66m versus 1.32m), significantly shorter stance phase percentage (60.3% versus 63.0%), and a significantly greater swing phase percentage (39.7% versus 37.0%) (44). The decreased stride length should be expected however, due to the depth of the steps in the study (0.26m).

Joint angles have been found to be significantly different between ascent and descent (54), as well as between stair ascent and level walking (44). This is logical because these processes would require different motions at the hip, knee, and ankle. For example, Protopapadaki et al. (54), whose SOI was the second step, found that there was significantly greater peak hip flexion during stair ascent (65.06˚) compared to descent (39.96˚). Also at the hip, Nadeau et al. (44), whose SOI was the first and second steps, found that there was a significant difference between level walking and stair ascent peak hip angles. During stair climbing, participants had a peak hip flexion angle of 60.1˚ and a peak hip extension angle of -4.7˚, each significantly different than level walking, which had an average peak of 30.8˚ and 15.5˚ of hip flexion and extension, respectively (44).

At the knee, Protopapadaki et al. (54) found decreased peak knee flexion during stair descent (90.52˚) compared to ascent (93.92˚). Nadeau et al. (44) found that the knee experienced an increased peak adduction angle during stair ascent (10.4˚) versus level walking (4.6˚). Additionally, this study found that the knee had a significantly increased peak knee flexion angle during stair ascent (93.1˚) compared to level walking (67.0˚), and a significantly increased peak knee extension angle during stair ascent (-10.0˚) versus level walking (1.1˚) (44).

The ankle also has significant differences between stair ascent and descent (54) and between stair ascent and level walking (44). Protopapadaki et al. (54) found that peak ankle dorsiflexion during ascent (11.2˚) was significantly lower than during descent (21.1˚), and that peak ankle plantarflexion was also significantly lower during ascent (31.3˚) than during descent (40.1˚). Compared to level walking, Nadeau et al. (44) found a significantly higher peak dorsiflexion angle (29.8˚) during stair ascent, versus 19.1˚ during level walking. This study also found that the peak ankle adduction angle while ascending stairs, 14.3˚, was significantly higher than during level walking (9.0˚) (44).

Many studies that have researched stair negotiation have looked closely at the effects of stair ascent and descent on joint kinetics, and GRF. This section of the review will focus on the effects of stair ascent and descent on GRFs as well as hip, knee, and ankle joint moments. Protopapadaki et al. (54) and Hamel et al. (26) studied the changes in GRFs when ascending and descending stairs. It was found that the first peak vertical GRF was higher in stair descent (26, 54) compared to ascent. For their young population (24.2±2.5 years), Hamel et al. (26) reported a first peak vertical GRF of 1.4 body weights (BW) during descent, compared to 1.2BW during ascent, a significant difference. In comparison, the second peak vertical GRF was higher during stair ascent than stair descent (26, 54). Hamel et al. (26) reported a significant different between the second peak vertical GRF during ascent (1.2BW) compared to descent (0.9BW). The increase in the first peak GRF during descent may be caused by participants increasing the impact force in order to slow down their descent in opposition to gravity, causing the first peak to be higher than the second peak during stair descent. During ascent, participants would increase the vertical force to counteract the force of gravity, causing the second peak to be higher than the first peak during stair ascent.

At the hip, Protopapadaki et al. (54) found that the peak external hip flexion moment was significantly higher during stair ascent (0.8Nm/kg) compared to stair descent (0.5Nm/kg). This study mentioned that the external hip moment was positive during most of stance phase in both ascent and descent conditions, and therefore a flexion moment, besides a brief period near the middle of stance during descent where there was a negative, extension hip moment. Nadeau et al. (44) found that the peak hip flexion moment during ascent (0.3Nm/kg) was significantly lower than during level walking (0.7Nm/kg), and that the peak external hip extension moment during stair ascent (0.5Nm/kg) was also significantly lower than during level walking (0.7Nm/kg). There is a noticeably large difference between the peak hip flexion moments during stair ascent between Protopapadaki et al. (54) (0.8Nm/kg) and Nadeau et al. (44) (0.3Nm/kg). This could be due to the age differences between the studies, 29.1 year old average (54) and 53.0 year old average (44).

Costigan et al. (13) who had a subject group with an average age of 24.6 years old, found that participants had a peak hip flexion moment of 0.8Nm/kg during stair ascent and 1.0Nm/kg during level walking. Additionally, this study found that the peak external hip adduction moment during stair ascent (0.8Nm/kg) was lower than during level walking (1.1Nm/kg), and that the peak internal rotation moment during stair ascent (0.3Nm/kg) was lower than during level walking (0.2Nm/kg) (13). Unfortunately, this study did not use any statistical analysis, so it is unknown if any of these differences are significant.

At the knee, Protopapadaki et al. (54) determined that stair ascent lead to a significantly higher peak knee extension moment (0.6Nm/kg) compared to stair descent (0.4Nm/kg). When comparing stair ascent to level walking, Nadeau et al. (44) found that the knee experienced a significantly higher peak extension moment during stair ascent (1.0Nm/kg) compared to level walking (0.5Nm/kg). Additionally, this study found that participants experienced a significantly higher first peak KAbM during stair ascent (0.8Nm/kg) compared to level walking (0.6Nm/kg). Costigan et al. (13) found that participants had a lower peak knee flexion moment during stair ascent (0.4Nm/kg) compared to level walking (0.5Nm/kg), a higher peak knee external adduction moment during stair ascent (1.2Nm/kg) compared to level walking, and a lower peak internal rotation moment at the knee during stair ascent (0.1Nm/kg) compared to level walking (0.1Nm/kg). However, as stated previously, Costigan et al. (13) did not report any significance for their values.

At the ankle, a study by Protopapadaki et al. (54) reported that the peak ankle dorsiflexion moment was, non-significantly, lower during stair descent (1.38Nm/kg) compared to ascent (1.5Nm/kg). Nadeau et al. (44) found that participants had a significantly lower peak dorsiflexion moment during stair ascent (1.2Nm/kg) compared to level walking (1.4Nm/kg). Standifird et al. (65) reported peak plantarflexion moments and found that during ascent the average first peak plantarflexion moment was -0.7±0.3Nm/kg, and the second peak plantarflexion moment had an average value of -1.1±0.1Nm/kg.

This literature review found that participants ascend stairs slower than they descend and walk on level ground (44, 54). Kinematically, this review found that participants have: a greater peak hip flexion angle during ascent compared to descent (44, 54), a lower peak hip extension angle during stair ascent compared to level walking (44), a decreased peak knee flexion angle during stair descent compared to stair ascent (54), an increased peak knee adduction and flexion angle during stair ascent compared to level walking (44), a decreased peak knee extension angle during stair ascent compared to level walking (44), a decreased peak ankle dorsiflexion angle during ascent compared to descent (54), a significantly higher peak dorsiflexion angle during stair ascent compared to level walking (44), and an increased peak ankle adduction angle while ascending stairs compared to level walking (44). Kinetically, this review found that participants have: a higher first peak vertical GRF during descent (26, 54), a higher second peak vertical GRF during stair ascent (26, 54), an increased peak external hip flexion moment during stair descent (54), a decreased peak external hip flexion moment during compared to level walking (44), a decreased peak external hip extension moment during stair ascent compared to level walking (44), an increased peak knee extension moment during stair ascent compared to descent (54), an increased peak knee extension moment during stair compared to level walking (44), an increased peak KAbM during stair ascent compared to level walking (44), and a decreased peak plantarflexion moment during stair ascent level walking (44).

*Stair Negotiation: Effects on Obese Participants*

To our knowledge, only one study has been published researching the effects of obesity on stair gait. This study by Strutzenberger et al. (68) focused on these effects in children (10.4±1.5years). Although it would be logical to assume that these effects would be different for children than adults, a study by Ganley and Powers (25) found that children that were only seven years old demonstrated the same lower extremity joint kinematics and kinetics, with the exception of ankle kinetics, compared to adults (31.8±6.8 years) while walking over level ground at an average velocity of 1.3±0.1 m/s. The children demonstrated significant differences in ankle kinetics from the adult group; the children in the study had a significantly lower peak plantarflexor moment (1.15Nm/kg) than the adults (1.56Nm/kg), and had significantly lower power absorption (-0.6±0.3W/kg) and generation (2.79±0.44W/kg) at the ankle during late stance than adults (-1.1±0.2W/kg and 3.5±0.6W/kg; absorption and generation, respectively). (25). This is in agreement to Cupp et al. (14), who showed that children above seven years old have similar kinetic patterns as adults (18-21 years), during level over ground walking at 1.4m/s and 1.2m/s, respectively. Any biomechanical differences found in children compared to adults during level walking should be amplified during stair gait due to the fixed step heights and stair ambulation demands.

The participants in the study by Strutzenberger et al. (68) were required to ascend and descend the stairs at a cadence of 110 steps/min. The stairs included six steps, with force plates in steps three and four for kinetic variable computation. The study found that the obese participants spent significantly less time in single support (0.39s) compared to healthy weight participants (0.42s) during stair ascent. During descent, obese participants spent significantly longer time in double support (0.27s) compared to healthy weight (0.24s). This longer time in double support may have been closely linked to the delay of toe-off from the step during descent. Obese participants tended to have toe-off at 63.3%, opposed to healthy weight participants who toed off at 61.2% (68). Kinematically, this study found that obese participants had a significantly greater minimum pelvis anterior tilt (20.1˚) compared to healthy weight participants (16.2˚) and a significantly greater peak knee abduction angle (-12.9˚) compared to healthy weight participants (-6.7˚) during stair ascent. Kinetically, they found that during stair descent, obese participants had a significantly lower peak hip extension moment (0.2Nm/kg) compared to healthy weight participants (0.4Nm/kg), and a significantly greater peak hip flexion moment (-0.5Nm/kg) compared to healthy weight participants (-0.4Nm/kg) (68). Additionally, obese participants during stair ascent had a significantly higher peak hip abduction moment (0.7Nm/kg) than healthy weight participants (0.6Nm/kg), and a significantly greater peak knee extension moment (1.1Nm/kg) compared to healthy weight participants (0.9Nm/kg) (68). No ankle kinetic variables had significant differences.

**Adjustments to Segment Inertia Parameters**

In biomechanics, it is important that we accurately portray body segment parameters so that kinetic variables can be correctly estimated. Typical parameters include mass, center of mass (COM) position, principle radii of gyration, and moment of inertia (16). These parameters are typically found through cadaver studies similar to Clauser et al. (11), adjusted later by Hinrichs (28), and Dempster (19), which is the most popular model currently used. Parameters found using these cadaver studies have been found to have large errors when used for calculating COM positions in healthy young male and female athletes (15). These errors were reduced by Zatsiorsky et al. (75) who used a gamma-ray scanner to determine the parameters of healthy and living young adults. According to de Leva (16), it is likely that the results from this study are not widely used because they used bony landmarks as reference points for locating segment COM and length.

de Leva (16) made adjustments to the mean relative CM positions and radii of gyration found by Zatsiorsky et al. (75). This was done in order to reference them to joint center positions, rather than bony landmarks. The first adjustment made by de Leva (16) was to calculate segment lengths. This was done by applying two equations, from Zatsiorsky et al. (75), and de Lava's personal communication with Zatsiorsky, twice, once about the sagittal axis and once about the transverse axis. The first equation was to find the mean length of the segment, which was calculated as: $T = \bar{r}_{abs}/\bar{r}_{rel}$ where T is the mean segment length, $\bar{r}_{abs}$ is the mean absolute radius of gyration of that segment about a given axis, and $\bar{r}_{rel}$ is the respective mean ratio between segment radius of gyration and length and was identical for all participants. For each segment, $\bar{r}_{abs}$ is estimated from equation two: $\bar{r}_{abs} = \sqrt{\bar{I}/\bar{m}}$, where $\bar{I}$ is the mean segment moment of inertia about the axis (sagittal or transverse), and $\bar{m}$ is the mean segment of mass (16). The results were then averaged and reported as the differences between segment length estimations, which ranged from 0.0 mm to 2.1 mm.

Another study by de Leva (17) reported the percent longitudinal distances of joint centers from neighboring bony landmarks. The model assumes that the joint centers lay on the segments longitudinal axis. de Leva (16) used these percentages to determine the segment lengths of the participants in the study by Zatsiorsky et al. (75). A graphic representation was given of the adjustments made to the segment COM positions, with the main differences being that they were estimated from a percent distance from a joint center, rather than a boney landmark. After the segment COM positions were adjusted, an equation was provided to find the segment moment of inertia (I). This equation is: $I = (M \times \bar{m}) \times (l \times \bar{r})^2$, where M is the participants mass, $\bar{m}$ is the mean relative segment mass, $l$ is the segment length, and $\bar{r}$ is the mean relative radius of gyration of the segment about the considered axis (16). The method by de Leva (16) provides useful method to adjust anthropometric parameters in order to improve the accuracy of joint kinetics calculations. Additional geometric models for segment COM and inertial parameter estimation. DeVita and Hortobagyi (20)used a method to make adjustments on these parameters based on research by Hanavan (27).

# Chapter III

## Materials and Methods

### Participants

For this study, healthy adults (18-40 years) were recruited through email and word of mouth. Participants who met the inclusion criteria for inclusion for either the healthy-weight or the obese group (Table 1) were asked to participate in the study. An *a priori* power analysis, using results from previous research (8, 20, 32, 44, 48, 49, 54), indicated that a total of 6-34 participants were needed for an *alpha* of 0.05 and a *beta* of 0.80. Variables used in the power analysis included first and second peak KAbM (44, 48, 49), knee extension moment (8, 20, 44, 54), and knee adduction/flexion/extension angles (32, 44). Due to a low number of published studies regarding obesity biomechanics, we expanded research the power analyses to include additional variables, aside from our primary variable (KAbM). All participants completed a physical activity readiness questionnaire (PAR-Q) (40) and signed an informed consent document approved by the Institutional Review Board of the University of Tennessee, Knoxville.

### Instrumentation

For three-dimensional (3D) kinematic data collection, a nine-camera motion analysis system (120Hz, Vicon Motion Analysis Inc., Oxford, UK) was used during testing. Participants wore tight-fitting spandex shorts and a t-shirt, as well as standardized running shoes (Noveto, Adidas, USA). Retroreflective anatomical markers were placed bilaterally on the 1$^{st}$ and 5$^{th}$ metatarsal heads, the distal end of the 2$^{nd}$ toe, medial and lateral aspects of the malleoli and femoral epicondyles, greater trochanters, iliac crests, and acromion processes. A semi-rigid thermoplastic shell with four retroreflective tracking markers was placed on the posterolateral aspects of each shank and thigh, as well as on the posterior trunk and top of the feet. Two additional shells, each with two tracking markers, were placed on the posterior-lateral aspect of the pelvis.

Two AMTI force platforms (1200 Hz, BP600600 and OR-6-7, American Mechanical Technology Inc., Watertown, MA, USA) were used to measure the ground reaction force (GRF) during level ground walking. An instrumented 3-step staircase (FP-stairs, American Mechanical Technology Inc., Watertown, MA, USA) bolted to the AMTI force platforms was used to measure GRF in conjunction of the two force platforms (Figure 1). Two additional steps ($4^{th}$ and $5^{th}$) were used in conjunction with the instrumented steps to allow for continuous gait across the measurement zone. Each step had a rise of 17.8 cm, a width of 60.0 cm, and a depth of 29.9 cm. A handrail was available on the right hand side (during ascent) to prevent a loss of balance or fall. Gait speed was measured using two photocells (63501 IR, Lafayette Instrument Inc., IN, USA) and two electronic timers (54035A, Lafayette Instrument Inc., IN, USA). For level walking, the photocells were placed 3 meters apart on either side of the force platforms and at shoulder height. For the stair ascent trials, the photocells were placed at the $1^{st}$ and $4^{th}$ steps. Body fat percentage for each participant was measured via a Tanita Body Composition Analyzer (BF-350, Tanita Corporation of America Inc., Arlington Heights, IL, USA). The researcher was trained on the proper use of the bio-electrical impedance scale. This training included proper input of height, gender, gender type (athletic versus not athletic), as well as proper cleaning methods.

**Experimental Procedures**

Each participant completed five successful ascent trials under both preferred and wide SW conditions. The second step of the staircase was the step of interest (65, 66).

Figure 1. Staircase used during data collection. The three lower steps were instrumented and the two upper steps were provided to allow continuous gait across instrumented section.

For consistency, only the right leg was tested. Participants were instructed to ascend the stairs so their right foot always landed on the second step.

Before trial data was collected the subject was asked to perform one static trial and one dynamic range of motion trial. A labeling skeleton template that includes both anatomical and tracking markers was attached to the participants file in Vicon Nexus 2.3 (Vicon Motion Analysis Inc., Oxford, UK) during the static trial. The static trial consisted of the participants standing on the third step of the stair case, with their arms folded across their chest. Approximately three seconds of data was recorded while the subject stood as motionless as possible. The markers were then labeled as the landmarks they represent on the participant. A static calibration was run to build a model of the participant within the computer system. This model is used to determine the individual subjects segment parameters, such has the length between the greater trochanter and lateral epicondyle. Once the model was built in the computer, the anatomical markers were removed, leaving the tracking markers. The labeling template was then changed to a template that included only tracking markers. The subject then performed the dynamic range of motion trial. This trial included 2-3 steps taken prior to ascending the staircase, ascent of the staircase, descent of the staircase, and 2-3 steps away from the staircase. The researcher then labeled the markers in the computer that correlated with those on the participant. A dynamic range of motion calibration was then run. This calibration helps the researcher by allowing the software a foreknowledge of what type of movements were to be expected during the data collection trials. This greatly improves the auto-labeling feature available to the researcher, and greatly reduces the amount of time spent labeling and gap filling the trials both while the participant is present and after the data collection has ended.

Prior to the stair ascent trials, participants practiced ascending the staircase at their self-selected (preferred) speed (3-5 trials). Practice trials were used to calculate average ascent speed as well as average referred step width. Data from the practice trials were exported to Visual3D biomechanical analysis software (C-Motion, Inc., Germantown, MD, USA). Preferred SW was calculated by finding the mediolateral distance between the center of mass of each foot during midstance while the subjects was on the second and third steps. All trials were performed within a speed range defined by the participant’s average speed during practice trials ± 5%. If the participant was unable to achieve this desired speed, the trial was performed again.

For the wide SW condition, SW was set at twice the participant’s preferred ascent SW observed during the practice trials. Doubling the preferred SW ensured that the wide SW condition was statistically larger than the preferred SW condition (48, 49). This manipulation was performed so that procedures resulted in reliably different gait behavior.

Colored markers were placed on each of the five steps to provide guides for foot placement. Participants were instructed to cover the marker with their mid-foot to achieve the desired SW. Targeting was prevented by the researcher giving verbal feedback to the participant. This was done so that the subject could walk normally, without leaning forward to look down at their foot placement. While it was expected that the participant would look at the steps as they would when they ascend a staircase in their day-to-day life, the researcher wanted to prevent any excessive leaning of the participant. Leaning forward, more than usual, while ascending the staircase would have likely altered their gait pattern, as well as caused the electronic timing gait to be triggered prematurely. Verbal instructions on any change of speed or step width were given to the participants by the researcher. Additionally, either the researcher or trained lab personnel helped to guide the subjects step width by giving verbal feedback while standing behind the participant, outside of the capture volume. Once it was believed that the participant had completed five trials at the correct speed and SW, the trials were batch processed in order to quickly label the tracking markers, as well as fill gaps in the marker trajectories, and then exported in to Visual 3D software (C-Motion, Inc., Germantown, MD, USA), where the researcher calculated the SW that was obtained during the trials. Any trials that did not meet the required SW, were repeated. The additional repetitions varied by participant, no participants performed more than five additional trials. The testing order of preferred and wide SW conditions was randomized using a random number generator in MATLAB (MATLAB, The MathWorks, Inc., Natick, Massachusetts, USA). Condition order favored preferred SW as the first condition to be tested with 14 participants completing the preferred SW first, and 10 participants completing the wide SW condition first (Appendix G, Table 16).

**Data Treatment and Analyses**

Raw data collected included 3-dimensional GRFs as well as marker coordinate data. The marker coordinate data was first analyzed in the Vicon Nexus 2.3 (Vicon Motion Analysis Inc., Oxford, UK). These data were checked for correct labeling throughout each trial. If any gaps were found, which was any point that a minimum of two cameras could not identify the marker position, the researcher filled the gaps using either a ridged body fill (if a minimum of three markers on the same shell as the missing marker were present throughout the gap) or a pattern fill (if less than three markers were available during the gap). Once all markers were labeled and gaps were filled, the researcher removed any “ghost” markers that appeared. A “ghost” marker occurs when a camera detects a reflection that is relayed as a possible marker, but no marker exists in the location identified by the camera. “Ghost” markers may can be caused by reflective surfaces in the laboratory, reflective clothing worn by those outside of the capture volume, or due to camera error.

After these errors were removed, data were exported into Visual3D biomechanical analysis software (C-Motion, Inc., Germantown, MD, USA) for 3D kinematic and kinetic computations. Angular computations were completed using a Cardan rotational sequence (X-Y-Z) and a right-hand rule to define angular variable conventions. Positive values indicated ankle dorsiflexion and inversion, knee extension and adduction, and hip flexion and adduction angles or moments. Kinematic and GRF data were filtered using a fourth-order Butterworth low-pass filter at 8 Hz for joint moment calculations; raw GRF data was filtered separately at 50Hz (45, 49). Filtering the data allows for removing noise in the data, generating smoother lines. The GRF was filtered at a higher frequency because when a lower frequency is used, the smoothed lines may hide a true maximum or minimum value. Two customized computer programs (VB_V3D and VB_Table, MS Visual Basic) determined variables of interest and organized data from Visual3D outputs for subsequent statistical analysis.

VB_V3D was used to identify points of interest during stance phase. These points included loading response (LR) and push-off response (PO) peak vertical and mediolateral GRF (filtered at 50Hz), LR peak knee extension moment, LR peak KAbM, PO peak KAdM, LR peak knee external rotation moment, PO peak knee internal rotation moment, as well as 3-dimensional peak angles (including both maximum and minimum angles, as well as contact and push-off angles). These events were picked for each trial by the researcher to ensure accuracy and consistency of the events. LR peak GRFs and moments were viewed as the loading of the body weight onto the step. These events typically occurred close to 25% of stance phase, and were followed by an unloading phase. PO peaks occurred later in stance (approximately 75-90%) and included a peak followed by a change in direction. The selected variables were then organized and saved in a separate Excel file for each participant. These Excel files organized each variable into a separate sheet. Each sheet included X, Y, and Z values for each of the five trials for the selected variables, as well as a mean and standard deviation for the five trials. The VB_Table program was used to organize and compute mean values for each participant, with variables organized into separate sheets of an Excel file. This program also generated the overall mean for each variable, along with means for each variable within the two groups.

GRFs and joint moments were normalized to lean body mass. Lean body mass was calculated by multiplying the body fat percent found through bio-electrical impedance by the total mass of the participant, and then subtracting this number from the participant's total mass.

**Statistical Analyses**

The data were first analyzed for any significant normality issues. No variables displayed significant non-normal distribution. However, high kurtosis was found for peak KAdM in healthy-weight participants during the preferred SW condition with a kurtosis value of 3.049 ($p=0.007$). It was determined that the likely cause of kurtosis was a small sample size, and that normal distribution could be assumed for all variables. A two-way (Group x SW) mixed design analysis of variance (ANOVA) was performed to analyze the main effects of group and SW, and to determine if any significant interactions were present during stair ascent (22.0 IBM SPSS, Chicago, IL). No violations of sphericity were found using the Mauchley's Test. The lack of sphericity in this test is due to the number of levels associated with the repeated measures ANOVA. Our study had only two levels (preferred and wide SW), while this test requires a minimum of three levels to find sphericity violations. When an interaction was revealed, a post-hoc comparison with Bonferroni adjustments was used to detect differences between groups and step width conditions. Independent samples t-tests were run to identify group differences in age, height, leg length, body fat percentage, and BMI. An *a priori* alpha level was set to 0.05.

# Chapter IV

## Effects of Wider Step Width on Knee Biomechanics in Obese and Healthy-Weight Participants During Stair Ascent

**Abstract**

An increased likelihood of developing obesity-related knee osteoarthritis may be associated with increased peak internal knee abduction moment. Increases in step width may act to reduce this moment. This study focused on how step width influenced the knee joint during stair ascent by healthy and obese participants. Participants ascended stairs while walking at their preferred speed and under one of two step width conditions – preferred and increased. Obese participants experienced greater mediolateral and vertical ground reaction forces (GRFs), as well as increased peak knee extensor moments and push-off peak internal knee adduction moments. The findings of this study indicate that when step width increases, obese participants will experience a disproportionate increase in Loading-response and push-off response peak mediolateral GRF, push-off peak knee adduction moments, and peak knee adduction angle compared to healthy participants. When normalized to lean body mass, obese participants also had greater increases in peak knee extension moments under the increased step width condition. Participants in each group experienced decreased in loading-response peak vertical GRF, loading-response peak knee abduction moment, peak knee internal rotation moment, knee extension range of motion, and knee abduction range of motion, and increased loading-response and push-off response peak mediolateral GRF, push-off peak knee adduction moment, peak knee external rotation moment, peak knee abduction angle, and knee internal rotation range of motion. This study provides important information regarding differences in knee joint biomechanics during stair ascent between obese and healthy populations.

## Introduction

Obesity is defined as having excessive body fat, leading to a BMI greater than 30 $kg/m^2$ (31), and has been associated with increased risk for knee osteoarthritis (OA), high blood pressure, dyslipidemia, heart disease, type 2 diabetes, respiratory dysfunction, and cancer, as well as other physiological conditions (3, 5, 12, 52). Research has found that individuals with BMIs greater than 30 are at risk for limitations in common activities of daily living, e.g., walking over level ground or stairs (3). Knee OA is likely a cause of some of these limitations, as obese participants are 6.8 times more likely to develop knee OA than healthy weight participants (12).

The development (5, 12) and progression (10) of medial compartment knee OA has been linked to increased medial compartment knee loading, and typically associated with an increased internal knee abduction moment (KAbM). Studies on level walking have shown that obese participants display an increase in non-normalized peak internal knee extension moments (6, 8) and KAbMs (6), compared to healthy weight individuals. When normalized to body mass, the difference in peak knee extension moments disappeared (6, 8). In contrast, the peak KAbM was found significantly lower in obese subjects when normalized to body mass (6). The normalization process masks the actual knee joint loading experienced during gait. The literature seems to suggest that the effects of obesity on knee joint loading become less clear when moments are normalized and therefore providing support for using non-normalized moments in obesity research.

No research has been performed on the effects of obesity on the knee joint during stair ascent. Stair ascent studies have found that healthy weight participants exhibit greater peak body mass normalized knee extension moments (44, 63) and decreased KAbMs (63), compared to level walking. Healthy-weight participants have also shown increased peak knee extension angles, knee extension ROMs, peak knee abduction angles, and knee abduction ROMs during stair ascent(44, 63)

Many researchers have employed various gait modifications in an attempt to reduce the peak KAbM. One prevalent idea is to increase the step width (SW) of a participant. A wider SW has been shown to reduce the peak knee extension moment and KAbM during level walking (4, 24, 76). This reduction in peak KAbM has also been demonstrated in stair negotiation in healthy (4, 49) and osteoarthritic (48) populations using wider SW. In these stair negotiation studies (47-49), the wide and wider SWs were set as 26% and 39% of the participant's leg length, respectively, based on previous work showing that healthy participants tend to walk on level ground at a SW that is 13% of their leg length, measured as the participant height up to the greater trochanter (38).

Previous studies have also demonstrated that obese participants tend to walk with an increased (SW), compared to healthy participants during level walking (8, 59, 62). These studies found that obese subjects walked at a SW of 0.15m (8) to 0.16m (62). The later found that the obese individuals walk with as much as a 100% increase in SW (from 8cm to 16cm), over healthy weight individuals (62). Other studies have found a more modest increase in SW of 30% (8) to 42% (59). Currently, it is unknown if obese individuals ascend stairs at a significantly greater SW. Additionally, it is unknown if an increase in SW will affect medial compartment knee joint loading similarly in the obese population as it does in the healthy weight population. Therefore, the purpose of this study was to determine the effects of increased SW and obesity on knee biomechanics during stair ascent. It was hypothesized that obese participants would experience higher peak KAbMs and knee extension moments than that healthy participants.

Additionally, it was hypothesized that an increase in SW would lead to decreased in peak KAbMs in both groups.

**Materials and Methods**

*Participants*

Fourteen healthy weight (age: 21.6±0.5years, height: 1.7±0.1m, weight: 66.3±9.3kg, BMI 22.5±1.9kg/m$^2$) and ten obese (age: 25.7±5.8years, height: 1.7±0.1m, weight: 100.6±12.6kg, BMI 32.8±2.7kg/m$^2$) participants were recruited to participate in the study (Table 2). Participants who met the criteria for inclusion in either the healthy weight group or the obese group, were asked to participate in the study. Inclusion criteria for this study were between 18 and 40 years old with a BMI between 30kg/m$^2$ and 39.9kg/m$^2$ for obese participants and between 18.0kg/m$^2$ and 24.9kg/m$^2$ for heathy weight participants. Exclusion criteria included any major lower extremity injuries or surgeries, any disorder/disease/pathology affecting gait or balance, and any lower extremity injuries within the past year (Table 1).

An *a priori* power analysis, using results from previous research (8, 20, 32, 44, 48, 49, 54), indicated that a total of 6-34 participants were needed for an *alpha* of 0.05 and a *beta* of 0.80.Variables used in the power analysis included first and second peak KAbM (44, 48, 49), knee extension moment (8, 20, 44, 54), and knee adduction/flexion/extension angles (32, 44). All participants completed a physical activity readiness questionnaire (PAR-Q) (40) and signed an informed consent document approved by the Institutional Review Board of the University of Tennessee, Knoxville, prior to inclusion in this study.

*Instrumentation*

For three-dimensional (3D) kinematic data collection, a nine-camera motion analysis system (240Hz, Vicon Motion Analysis Inc., Oxford, UK) was used during testing. Participants were asked to wear tight-fitting spandex shorts and t-shirt, as well as standardized running shoes (Noveto, Adidas, USA). Retroreflective anatomical markers were placed bilaterally on the $1^{st}$ and $5^{th}$ metatarsal heads, the distal end of the $2^{nd}$ toe, medial and lateral aspects of the malleoli and femoral epicondyles, greater trochanters, iliac crests, and acromion processes. A semi-rigid thermoplastic shell with four reflective tracking markers was placed on the posterolateral aspects of each shank and thigh, as well as on the posterior trunk and in the center on the anterior side of the foot. Two additional shells, each with two tracking markers, were placed on the posterior-lateral aspect of the pelvis. Once a static trial was obtained, anatomical markers were removed before data collections of movement trials.

Two AMTI force platforms (1200 Hz, BP600600 and OR-6-7, American Mechanical Technology Inc., Watertown, MA, USA) were used to measure the ground reaction force (GRF). An instrumented 3-step staircase (FP-stairs, American Mechanical Technology Inc., Watertown, MA, USA) bolted to the AMTI force platforms, was used to measure GRF in conjunction of the two force platforms (Figure 1). Two additional steps ($4^{th}$ and $5^{th}$) were used in conjunction with the instrumented steps to allow for continued motion following the three instrumented steps. Each step had a rise of 17.8 cm, a width of 60.0 cm, and a depth of 29.9 cm. In order to prevent a loss of balance and/or a fall, a handrail was available on the right hand side during stair ascent. Speed of stair ascent trials was monitored by a set of two photo cells (63501 IR, Lafayette Instrument Inc., IN, USA) and two electronic timers (54035A, Lafayette Instrument Inc., IN, USA). The photo cells were set at the $1^{st}$ and $4^{th}$ steps. Body fat percentage for each subject was measured via a Tanita Body Composition Analyzer (BF-350, Tanita Corporation of America Inc., Arlington Heights, IL, USA).

*Experimental Procedures*

Each participant performed five successful trials of stair ascent with preferred SW and five successful trials of stair ascent with wide SW. The stair of interest was selected as the second step of the stair case during both ascent and descent conditions (65, 66). For consistency, the right leg was tested for each subject. Each subject was asked to ascend the stairs so that they used their right foot to step on the second step.

Prior to data collections, participants were asked to practice ascending the stair case at their self-selected (preferred) speed for three to five trials. These practice trials were also used to obtain an average walking speed and placement of the respective foot on the force platform or step without targeting. Data from the practice trials were exported to a Visual3D biomechanical analysis software suite (C-Motion, Inc., Germantown, MD, USA). Preferred step widths were calculated by finding the mediolateral distance between the center of masses of both feet during midstance. Step widths were calculated using the second and thirds steps. A speed range, average speeds ±5% of the practice trials, was used to monitor the movement trials.

For the wide step width conditions, the step widths were set at twice the participant's preferred step width, collected and calculated during the practice trials. Doubling the preferred step width ensured that our wide step width conditions were statistically larger than the preferred condition (48, 49).

Strips of masking tape were placed to each of the five steps of the staircase to mark the preferred and wide SW. Different colored markers were used to draw lines, running anteroposterior, on the masking tape. These lines were used for target guides for the participant's foot placement during stair negotiation. The participants were then asked to walk following the lines of tape, using their mid-foot to cover the line, in order to achieve the desired step width. A trial was deemed successful if the speed was within the participant's preferred gait speed range, the second step was contacted with the right foot at the correct SW, and without targeting. Targeting was prevented by the researcher giving verbal feedback to the participant. This was done so that the participant could walk normally, without leaning forward to look at their foot placement. Any unsuccessful trial was repeated. The testing order of the preferred and wide step widths were randomized by using a random number generator in MATLAB (MATLAB, The MathWorks, Inc., Natick, Massachusetts, USA). Testing order for each participant can be found in Appendix H, Table 17.

*Data Analysis*

The data collected were exported into a Visual3D biomechanical analysis software suite (C-Motion, Inc., Germantown, MD, USA) for 3D kinematic and kinetic computations. A Cardan rotation sequence (Sagittal-Frontal-Transverse, X-Y-Z) was used for 3D angular computations and a right-hand rule was used define angular kinematic and kinetic variable conventions. Positive values indicate ankle dorsiflexion and inversion, knee extension and adduction, and hip flexion and adduction angles or moments. Kinematic and GRF data were filtered using a fourth-order Butterworth low-pass filter at 8 Hz for joint moment calculations, raw GRF data was filtered a second time at 50Hz (45, 49). Two customized computer programs (VB_V3D and VB_Table, MS Visual Basic) were used to determine variables of interest during both loading response (LR, typically first 25% of stance as body weight is loaded onto the step) and push-off response (PO, typically occurs within last 25% of stance as participant pushes themselves up to the next step) and to organize data for statistical analysis from the outputs of Visual3D.

GRFs and joint moments were normalized by lean body mass normalization. This was done by first finding the lean body mass of each subject. Lean body mass was calculated by multiplying the body fat percent found through bio-electrical impedance by the total mass of the participant, and then subtracting this number from the participant's total mass. The GRF was then normalized by the calculated lean body mass.

*Statistical Analysis*

A two-way (Group x SW) mixed design analysis of variance (ANOVA) was performed to analyze the main effects of group and SW, and to determine if any significant interactions were present during stair ascent (22.0 IBM SPSS, Chicago, IL). An *a priori* alpha level was set to 0.05. Independent samples t-tests were run to identify group differences in age, height, leg length, body fat percentage, and BMI.

**Results**

No significant differences were found between groups for age, height, leg length, ascent speed, or preferred SW (Table 18). Body mass, body fat percentage, and body mass index (BMI) was higher in the obese group compared to the healthy-weight group ($p<0.001$, each). SW was wider during the wide ascent condition for both the healthy-weight and obese groups ($p<0.001$, each), compared to preferred SW (Table 2, Table 19). Results of independent samples t-tests, including F and p values for Levene's Test for Equality of Variances as well as t and p values for the t-tests can be found in Appendix F, Table 17. F, p, and $\eta^2$ values from the ANOVAs of reported variables can be found in Appendix H, Table 19

*Ground Reaction Force*

A step width main effect found decreases in LR vertical GRF ($p=0.045$). No Group x SW interaction was found for LR or PO peak vertical GRFs. When normalized to lean body mass, the LR vertical GRF was decreased by an increased SW ($p=0.038$) (Table 8).

Increased SW significantly increased the LR and PO peak mediolateral (ML) GRFs within each group (all $p<0.001$). Significant Group x SW interactions were observed for both LR and PO peak mediolateral (ML) GRFs (all $p<0.001$). This interaction indicated that both groups increased GRF due to the wide SW manipulation, but the obese group increased to a greater extent. The obese group had greater mean LR peak ML GRF values during both preferred ($p=0.008$) and wide ($p=0.001$) SW. Additionally, the obese group demonstrated greater mean PO peak values for preferred ($p=0.033$) and wide ($p=0.001$) SW. When normalized to lean body mass, increased SW continued to significantly increase the LR and PO ML GRF (all $p<0.001$). However, the interaction disappeared for the LR ML GRF (Table 8).

*Joint Kinetics*

No main effects ($p=0.091$) or interactions ($p=0.264$) were found for the peak knee extension moment (Table 19). When we normalized the peak knee extension moment to lean body mass, no SW effects or interactions were found.

Increased SW generated significant reductions in LR peak KAbM ($p=0.009$). An interaction that approached significance was discovered ($p=0.051$). Post-hoc comparisons revealed that only obese participants experienced a decrease in LR KAbM when SW was increased ($p=0.020$). Increased SW significantly increased PO peak KAdM ($p<0.001$). An interaction ($p=0.022$) was found for PO peak KAdM. Obese participants experienced a larger increase in KAdM, when SW was increased from preferred to wide, compared to healthy. Post-hoc comparisons showed increases in PO peak knee adduction moment (KAdM) for both healthy ($p<0.001$) and obese ($p=0.003$) participants when SW was increased. The obese group also exhibited an increased PO KAdM during the wide SW condition compared to the healthy group ($p=0.013$) (Table 5).

When normalized to lean body mass the wide SW condition continued to produce a significantly lower LR peak KAbM ($p=0.031$) (Table 7). When normalized to lean body mass, the Group by SW interaction for the PO KAdM disappeared; however, it still approached significance level ($p=0.063$). The normalized KAdM still increased with the increased SW ($p<0.001$) (Table 7).

Increased SW increased LR peak knee external rotation moment ($p<0.001$) and decreased PO peak internal rotation moment ($p=0.008$) (Table 6). No Group x SW interactions were found in transverse plane knee kinetics. Similar SW effects where seen when normalized to lean body mass. LR peak knee external rotation moment was significantly increased ($p=0.001$), and PO peak knee internal rotation moment was significantly decreased ($p=0.004$) (Table 7)

*Knee Kinematics*

Increased SW did not affect peak knee extension angle, however, knee extension ROM was decreased ($p<0.001$) (Table 6). The frontal plane results revealed a reduction in peak knee abduction angle ($p<0.001$), as well as a significant Group x SW interaction for the peak abduction angle ($p=0.005$, Table 6). These interactions revealed that the increased SW generated a larger increase in the peak knee abduction angle for obese participants than it did for the healthy participants. Post-hoc tests showed a greater peak abduction angle for obese participants during both preferred ($p=0.040$) and wide ($p=0.002$) SW conditions compared to the healthy.

In addition, the wide SW condition produced a decrease in the peak knee abduction ROM ($p<0.001$). The Group x SW interaction was also significant for knee abduction ROM ($p=0.10$). Post-hoc tests revealed that an increased SW caused a reduced knee abduction ROM for both healthy ($p<0.001$) and obese ($p=0.023$). This ROM was also lower for healthy participants ($p=0.022$) compared to obese during the wide SW condition. Overall, obese subjects had a greater reduction in knee abduction ROM than healthy subjects. In the transverse plane, increased SW lowered the peak knee internal rotation angle (p=0.001) and knee internal rotation ROM (0.022).

**Discussion**

The purpose of this study was to determine the effects of SW on knee biomechanics of obese and healthy-weight participants during stair ascent. Our first hypothesis that increasing SW would significantly reduce KAbM during this task was supported. The SW main effect for the LR KAbM demonstrated that increased SW had a reduction effect on the KAbM. Our findings indicated that the LR peak KAbM was reduced in obese participants when SW was increased. These results were consistent with previous research during level walking (4, 24, 76), as well as stair negotiation in healthy-weight (4, 49) and osteoarthritic (48) populations. In contrast, healthy-weight participants did not present a reduction in this moment, failing to replicate findings of previous studies. The presence of a reduced KAbM in obese participants when SW was increased may be due to obese subjects experiencing greater increases in the peak knee abduction angle when SW was increased, while healthy participants did not have a significant increase in peak knee abduction angle. Our results indicate that as peak knee abduction angle increases, the peak LR KAbM is reduced, during stair ascent. This is likely related to a reduction of the frontal plane moment arm of the GRF vector about the knee joint. No differences in leg length, SW, or walking speed were found between groups, and did not affect the KAbM between groups

We found that both healthy weight and obese subjects exhibited a push-off peak internal knee adduction moment (KAdM). These moments were significantly increased in the presence of a wide SW for both obese and healthy groups. The increase was more pronounced in those in the obese group, leading to a significant interaction effect. In contrast, previous studies on knee biomechanics during stair ascent have demonstrated that participants experience a second PO peak KAbM (4, 48, 49, 65), rather than a peak KAdM. Since our preferred SW (0.14±0.04 for healthy and 0.17±0.04 for obese) was similar to those previously found in healthy participants while ascending stairs (48), it is unknown why our study found such a different frontal plane knee moment pattern during ascent. We speculate that increasing SW may have the tendency to cause a shift in knee joint loading so that the medial compartment experiences less loading, at the cost of greater loading to the lateral compartment. Although groups experienced an increase in peak knee abduction angle, there was an overall reduction in abduction ROM, which may be related to the change in moment from abduction to adduction. In a similar way to how increasing the peak knee abduction angle decreases the LR peak KAbM, we argue that this reduction in knee adduction leads to the generation of higher PO peak KAdMs. In agreement with previously published reports, the ML GRF was significantly increased due to a wide SW (8, 38), this increase in ML GRF was significantly higher for obese participants, and may have also contributed to the differences found in front plane knee moments.

Our hypothesis that peak KAbMs would be greater in obese participants than in healthy participants was partially supported by our findings, this is because PO peaks were found to be significantly different, while LR peaks were not. This could be due to the increased peak knee abduction exhibited by obese subjects. As previously stated, increases in this angle seem to have an inverse relationship with the LR peak KAbM. However, it was found that obese subjects did exhibit greater PO peak KAdMs during the wide SW condition. It is interesting to note that both groups did not exhibit different PO peak KAdMs during the preferred SW condition. This is likely due to similar knee abduction ROMs between the two groups during the preferred SW condition, but significantly reduced abduction ROMs during the wide SW condition. In addition to this, obese subjects experienced much larger increases in peak knee abduction angles, while healthy participants experienced a greater reduction in knee abduction ROM, when SW was increased. This may have led to an increased moment arm more for of the obese participants than those of the healthy participants.

Our third hypothesis was that obese participants would generate higher peak knee extension moments than healthy participants. Our findings support our hypothesis. We found that obese subjects exhibited higher peak knee extension moments during both preferred and wide SW conditions. This result is expected as obese subjects also experienced higher peak vertical GRFs than healthy weight subjects, due to their significantly increased mass. This finding is in agreement with previous research in level walking (8). Other research on level walking found that this difference disappears when normalized to body mass (8), body mass and height (32), or lean body mass and height (23). DeVita and Hortobagyi (20) found that obese participants have lower peak knee extension moments when normalized to by mass, than healthy participants. Increasing SW had little to no effect (p=0.091) on the extension moment for either group.

Few studies have described transverse plane kinetics while ascending stairs. Both groups showed a peak external rotation moment during LR, and a peak internal rotation moment during PO. Obese participants demonstrated a higher peak LR external rotation moments than healthy subjects in both SW conditions. No group differences were found in the peak PO internal rotation moment. Increased SW had a significant effect on each of these moments. The peak LR external rotation moment was significantly increased for each group, while the PO peak internal rotation moment was decreased for the healthy group only. This may be due to a significantly increased peak knee internal rotation angle for the obese group, but not for the healthy group. To our knowledge, this is the first time changes in the knee about the transverse plane due to an increased SW has been reported for stair ascent. Lai et al. (32) did not find any significant differences, during level walking, in peak external or internal rotation moments between healthy and obese participants, when moments were normalized to body mass and height.

Many previous studies on obese and healthy participants report finding differences in SW. Namely, they all have the tendency to show obese participants having a wider preferred SW than healthy participants (8, 59, 62). Interestingly, our study did not find that these two groups had significant SWs during stair ascent. While obese participants tended to have a larger SW than healthy participants, this value was not significant at only 8.2% larger for preferred and 8.6% larger for wide SW conditions. Participants in our study were not given any instruction on SW during their preferred SW trials, and were free to walk at their preferred SW. Results from our study indicate that healthy participants ascended at an average of 16.4±3.6% and obese participants ascended at 19.3±5.3% (not significantly larger, p=0.181) of leg length. This is a similar percentage found by Paquette et al. (48), who found that healthy participants walked ascended stairs at 15.4±3.1% of leg length. Previous studies have shown that subjects tend to walk over level ground at a SW that is approximately 13% of their leg length measured as distance from the anterior superior iliac spine, to the medial malleolus (48, 49). Studies on stair ascent have used this 13% to establish wide SW conditions (48, 49).

Walking speed is also often reported as being significantly different between these two subject groups during level walking. Obese participants reportedly have a 0.3 m/s, or 16% (20), slower preferred walking velocity than healthy weight participants (18, 23, 32, 58, 60, 62). This reduction in walking speed is important because increases in walking speed cause increased vertical, anteroposterior, and mediolateral ground reaction forces (GRF) experienced by both healthy and obese participants (8). Additionally, increased walking speed increases peak hip flexion/extension moments, and peak knee extension/flexion moments (34). However, we found no differences in self-selected stair ascent speed between groups during either preferred (p=0.493) or wide (p=0.387) SWs for our stair ascent task.

Three limitations to this study should be noted. It is common knowledge that participants with excessive adipose tissue are much harder to palpate their bony landmarks than lean participants. This may lead to small errors in marker placement for the group. Additionally, this study did not take into account the soft tissue artifact. The soft tissue artifact may be higher in the obese population than in the healthy weight population due to the movement of underlying adipose tissue (29, 51). Although the number of participants in the obese group met the required minimum of estimated sample size and the observed power reached an acceptable level for significant differences that were found, it is still considered a small sample size. For group differences, our statistical analyses revealed that our observed power level reached acceptable levels ranging 0.777 – 1.00 for selected variables showing significant interaction and main effects including peak ML GRF, peak vertical GRF, KAbM, and peak knee extension moment. Finally, it is of importance to note that the use of a bio-electrical impedance scale is not the most reliable form of body fat estimation, as it is heavily reliant on proper use of the equipment as well as additional factors such as hydration level of the participant. While individual results may vary, research has shown that this type of body composition analysis will provide accurate group means, when compared to gold-standard type tests (i.e. dual x-ray absorptiometry) (55).

*Conclusion*

This study was performed to find differences of knee biomechanics in obese and healthy weight people during the stair ascent task, while ascending at their own preferred SW and the wider SW.

We found that Obese subjects experienced higher non normalized ML and vertical GRFs. With the exception of the peak KAbM, obese participants consistently experienced greater peak knee extension moments and peak KAdMs. The results also revealed that there were significant interactions that cause obese and healthy participants to have different changes in their knee biomechanics during ascent. This research also demonstrates the importance of analyzing non normalized GRF and knee moment data because we were able to reveal the extent in which obese subjects experience these loads. Further research should be performed to expose how SW effects biomechanics of the hips and ankles in obese and healthy-weight individuals.

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## Appendices

**Appendix A: Inclusion and Exclusion Criteria/Participant Demographics**

Table 1. Exclusion and Inclusion Criteria for Healthy-Weight and Obese Participants.

| Exclusion Criteria | Inclusion Criteria | |
|---|---|---|
| Healthy-Weight and Obese | Healthy-weight | Obese |
| Any major lower extremity injuries or surgeries | Age between 18-40 | Age between 18-40 |
| Any disorder/disease/pathology affecting gait or balance | BMI between 18.0 $kg/m^2$ and 24.9 $kg/m^2$ | BMI between 30.0 $kg/m^2$ and 39.9 $kg/m^2$ |
| Any lower extremity injuries within the past year | | |
| Any pain while performing common activities of daily living | | |
| Women who are pregnant or nursing | | |
| Any cardiovascular disease of primary risk factor indicated by the PAR-Q | | |

Table 2. Participant Demographics: mean ± STD.

| | Healthy | Obese | p |
|---|---|---|---|
| Age (years) | 21.64±0.50 | 25.70±5.79 | 0.054 |
| Height (m) | 1.71±0.08 | 1.75±0.06 | 0.199 |
| Mass (kg) | 66.28±9.31 | 100.60±12.61 | **<0.001** |
| Leg Length (m) | 0.84±0.05 | 0.86±0.03 | 0.217 |
| BFP (%) | 22.67±6.58 | 36.56±7.67 | **<0.001** |
| BMI (kg/m$^2$) | 22.53±1.85 | 32.79±2.66 | **<0.001** |
| Preferred Step Width (m) | 0.14±0.04[a] | 0.17±0.04[a] | **<0.001** |
| Wide Step Width (m) | 0.30±0.07 | 0.35±0.07 | |
| Preferred Step Width Speed (m/s) | 0.62±0.06 | 0.60±0.08 | 0.493 |
| Wide Step Width Speed (m/s) | 0.62±0.06 | 0.60±0.08 | 0.387 |

[a]: Significantly different between Step Widths of the same group.

BFP: Body Fat Percent

BMI: Body Mass Index

Table 3. Individual Subject Characteristics.

| Subject | Group | Gender | Age (yrs.) | Height (m) | Weight (kg) | BMI (kg/m$^2$) |
|---|---|---|---|---|---|---|
| 2 | Obese | Male | 31 | 1.86 | 123.2 | 35.80 |
| 4 | Healthy | Female | 22 | 1.80 | 80.1 | 24.86 |
| 5 | Healthy | Female | 22 | 1.61 | 53.5 | 20.77 |
| 9 | Healthy | Female | 22 | 1.65 | 57.5 | 21.25 |
| 10 | Healthy | Male | 22 | 1.66 | 56.5 | 20.63 |
| 11 | Healthy | Male | 22 | 1.75 | 62.8 | 20.51 |
| 12 | Obese | Male | 21 | 1.73 | 90.1 | 30.10 |
| 13 | Healthy | Female | 22 | 1.62 | 65.2 | 24.84 |
| 14 | Healthy | Female | 21 | 1.77 | 76.4 | 24.52 |
| 15 | Healthy | Male | 22 | 1.66 | 56.4 | 20.59 |
| 16 | Healthy | Male | 21 | 1.66 | 59.8 | 21.70 |
| 17 | Healthy | Female | 21 | 1.75 | 75.8 | 24.75 |
| 18 | Healthy | Male | 21 | 1.78 | 78.5 | 24.92 |
| 19 | Healthy | Female | 22 | 1.65 | 62.3 | 22.88 |
| 20 | Obese | Male | 21 | 1.65 | 86.8 | 31.88 |
| 21 | Healthy | Male | 22 | 1.89 | 75.6 | 21.16 |
| 22 | Obese | Female | 22 | 1.73 | 103.6 | 34.62 |
| 23 | Obese | Female | 22 | 1.73 | 89.8 | 30.00 |
| 24 | Healthy | Female | 21 | 1.75 | 67.5 | 22.04 |
| 25 | Obese | Male | 25 | 1.81 | 103.7 | 31.65 |
| 26 | Obese | Male | 29 | 1.73 | 93.1 | 31.29 |
| 27 | Obese | Female | 22 | 1.73 | 89.8 | 30.00 |
| 28 | Obese | Female | 25 | 1.75 | 110.9 | 36.42 |
| 29 | Obese | Male | 39 | 1.79 | 115 | 36.09 |

## Appendix B: Chapter IV Tables

Table 4. Peak Mediolateral and Vertical GRFs (N) for Stair Ascent: mean ± STD.

| Variable | Healthy Preferred SW | Healthy Wide SW | Obese Preferred SW | Obese Wide SW | Int. p | Grp. p | SW p |
|---|---|---|---|---|---|---|---|
| LR Peak Vertical GRF | 759.6±96.0[a,#] | 728.6±85.2[#] | 1079.2±106.5 | 1069.0±150.9 | 0.296 | **<0.001** | **0.045** |
| PO Peak Vertical GRF | 823.8±142.2[#] | 820.3±118.9[#] | 1136.8±111.0 | 1163.0±114.6 | 0.349 | **<0.001** | 0.472 |
| LR Peak ML GRF | 39.0±14.1[a, #] | 91.78±20.3[#] | 59.3±20.4[a] | 135.5±35.1 | **<0.001** | **0.002** | **<0.001** |
| PO Peak ML GRF | 29.7±17.3[a,#] | 82.2±24.9[#] | 48.9±24.5[a] | 131.6±40.1 | **<0.001** | **0.004** | **<0.001** |

[a]: Significantly different from Wide SW of the same subject group.

[#]: Significantly different from Obese of the same SW.

ML: Mediolateral

GRF: Ground Reaction Force

Int.: Interaction

Grp.: Group Main Effect

SW: Step Width

Table 5. Peak Knee Loading and Push-Off Response Moments for Stair Ascent (Nm): mean ± STD.

| Variable | Healthy Preferred SW | Healthy Wide SW | Obese Preferred SW | Obese Wide SW | Int. p | Grp. p | SW p |
|---|---|---|---|---|---|---|---|
| LR Knee Extension Moment | 104.1±22.6[#] | 105.3±25.2[#] | 153.8±26.3 | 159.7±29.4 | 0.264 | **<0.001** | 0.091 |
| LR Knee Abduction Moment | -21.8±11.1 | -20.7±7.7 | -25.0±11.3[a] | -18.5±14.2 | 0.051 | 0.904 | **0.009** |
| PO Knee Adduction Moment | 10.8±4.2[a] | 15.0±5.7[#] | 17.7±15.8[a] | 27.5±12.6 | **0.022** | **0.022** | **<0.001** |
| LR Knee External Rotation Moment | 9.0±4.9[a,#] | 12.5±4.4[#] | 15.6±6.7[a] | 18.6±4.9 | 0.763 | **0.005** | **<0.001** |
| PO Knee Internal Rotation Moment | -8.9±1.8[a] | -7.2±1.1 | -10.1±4.5 | -9.3±4.7 | 0.258 | 0.168 | **0.008** |

[a]: Significantly different from Wide SW of the same subject group.

[#]: Significantly different from Obese of the same SW.

LR: Loading Response

PO: Push-Off

Int.: Interaction

Grp.: Group Main Effect

SW: Step Width

Table 6. Knee Contact, Peak, and ROM Angles for Stair Ascent (degrees): mean ± STD.

| Variable | Healthy Preferred SW | Healthy Wide SW | Obese Preferred SW | Obese Wide SW | Int. p | Grp. p | SW p |
|---|---|---|---|---|---|---|---|
| Peak Knee Flexion Angle | -8.6±4.4 | -9.4±5.9 | -10.4±4.7 | -10.3±4.5 | 0.340 | 0.530 | 0.489 |
| Knee Extension ROM | 58.1±4.8[a] | 54.8±6.0 | 55.0±7.1[a] | 51.9±7.9 | 0.895 | 0.252 | **<0.001** |
| Peak Knee Abduction Angle | -1.6±2.5[#] | -2.1±2.3[#] | -3.8±2.4[a] | -5.8±2.8 | **0.005** | **0.007** | **<0.001** |
| Knee Abduction ROM | -12.6±3.8[a] | -9.0±3.3[#] | -14.2±3.5[a] | -12.8±4.1 | **0.010** | 0.086 | **<0.001** |
| Peak Knee Internal Rotation Angle | -10.2±2.9 | -9.8±3.1 | -9.4±4.1 | -8.2±4.4 | 0.416 | 0.407 | 0.101 |
| Knee Internal Rotation ROM | -9.3±4.3 | -10.9±3.7 | -8.9±4.5[a] | -10.4±5.7 | 0.465 | 0.505 | **0.022** |

[a]: Significantly different from Wide SW of the same subject group.

[#]: Significantly different from Obese of the same SW.

ROM: Range of Motion

Int.: Interaction

Grp.: Group Main Effect

SW: Step Width

Table 7. Lean Body Mass Normalized Peak Knee Loading and Push-Off Response Moments for Stair Ascent (Nm): mean ± STD.

| Variable | Healthy Preferred SW | Healthy Wide SW | Obese Preferred SW | Obese Wide SW | Int. p | Grp. p | SW p |
|---|---|---|---|---|---|---|---|
| LR Knee Extension Moment | 2.033±0.327[#] | 2.050±0.336[#] | 2.429±0.303 | 2.513±0.277 | 0.363 | **0.002** | 0.178 |
| LR Knee Abduction Moment | -0.421±0.175 | -0.410±0.132 | -0.398±0.170[a] | -0.295±0.234 | 0.073 | 0.329 | **0.031** |
| PO Knee Adduction Moment | 0.217±0.094[a] | 0.303±0.124 | 0.300±0.278[a] | 0.455±0.243 | 0.063 | 0.133 | **<0.001** |

[a]: Significantly different from Wide SW of the same subject group.
[#]: Significantly different from Obese of the same SW.
LR: Loading Response
PO: Push-Off
Int.: Interaction
Grp.: Group Main Effect
SW: Step Width

Table 8. Peak Mediolateral and Vertical GRFs (Lean Body Mass) for Stair Ascent: mean ± STD.

| Variable | Healthy Preferred SW | Healthy Wide SW | Obese Preferred SW | Obese Wide SW | Int. p | Grp. p | SW p |
|---|---|---|---|---|---|---|---|
| LR Peak Vertical GRF | 1.53±0.17[a,#] | 1.47±0.17[#] | 1.75±0.20 | 1.73±0.27 | 0.216 | **0.006** | **0.038** |
| PO Peak Vertical GRF | 1.65±0.24 | 1.65±0.22[#] | 1.85±0.24 | 1.90±0.31 | 0.373 | **0.036** | 0.417 |
| LR Peak ML GRF | 0.08±0.02[a,] | 0.18±0.03 | 0.09±0.03[a] | 0.22±0.05 | 0.114 | 0.069 | **<0.001** |
| PO Peak ML GRF | 0.06±0.02[a] | 0.16±0.03[#] | 0.08±0.03[a] | 0.21±0.07 | **0.040** | **0.027** | **<0.001** |

[a]: Significantly different from Wide SW of the same subject group.
[#]: Significantly different from Obese of the same SW.
ML: Mediolateral
GRF: Ground Reaction Force
Int.: Interaction
Grp.: Group Main Effect
SW: Step Width

**Appendix C: Informed Consent**

**INFORMED CONSENT FORM**

Effects of Wider Step Width on Knee Biomechanics in Obese and Healthy-Weight Participants During Stair Ascent

Principal Investigator: Derek Yocum, B.S.
Address: 136 HPER
1914 Andy Holt Avenue
Knoxville, TN 37996
Phone: (865) 974-2091

Faculty Advisor: Songning Zhang, PhD
Address: 340 HPER
1914 Andy Holt Avenue
Knoxville, TN 37996
Phone: (865) 974-2091

**Introduction**

You are invited to participate in this research study because you are an adult between 18 and 40 years old. This research investigates the differences in knee function joint in both obese and normal weight people. Please ask the study staff to explain any words or information that you do not clearly understand. Before agreeing to participate in this study, it is important that you read and understand the following explanation of the procedures, risks, and benefits.

**Testing Protocol**

If you agree to participate, you will attend one study visit at the Biomechanics/Sports Medicine Lab on the UT campus. Your information from the demographic questionnaire and Physical Activity Readiness Questionnaire (PAR-Q), will be used for this study. The study visit will take approximately 2½ – 3 hours. You will need to wear clothing appropriate for exercise which includes spandex shorts and t-shirt. If you do not have spandex type of clothing, spandex short or laboratory paper short will be provided**.**

We will measure your weight and height. We will place reflective markers on your feet, ankles, legs, knees, thighs, pelvis and trunk. This will allow motion cameras to capture your movements when performing the exercises. The cameras will not record images of you. If you have any questions, interests, or concerns about any equipment to be used in this test, please feel free to ask the investigator or other research personnel.

You will perform the following exercises at your own speed:

- Walk across the floor 5 times using your normal stride.
- Walk across the floor 5 times using a wider stride than normal.
- Climb up and down stairs 5 times using your normal stride.
- Climb up and down stairs 5 times using a wider stride than normal.

Trials need to be completed at self-selected speed. You will be asked perform several practice trials for overground walking and stair negotiation to become familiar with the testing procedures and determine your self-selected speeds. During testing trials, you will be asked to walk within 5% of the average speed found during the practice trials. If you are not within 5%, you will be asked to repeat the trial. It is anticipated that you will not be required to perform more than ten to twelve trials of each test condition.

You can practice these exercises to familiarize yourself with the test procedures. You can take breaks as needed. You can end any exercise early and do not have to complete the study visit.

**Potential Risks**

Risks associated with this study are minimal. There is a small risk injury but it is no greater than the risk you experience when doing these activities on a daily basis. You can practice the exercises before the testing and take breaks as needed. If you are injured the study visit, we will provide standard first aid. In the unlikely event you are injured during the study, the University of Tennessee does not automatically provide reimbursement for medical care or other compensation and you will be responsible for any medical expenses. If you are injured, please notify Derek Yocum or his advisor, Dr. Songning Zhang (974-2091).

Every research study involves some risk to your confidentiality. It is possible that other people could find out you were in the study or see your study information. But we will do our best to keep your information confidential to minimize this risk.

**Benefits of Participation**

You may not benefit from participation in this study directly. However, you may learn about abnormalities that might be corrected with gait movement modifications, and footwear and orthotic choices. You can receive an individual report of your study results to share with your personal physician. Results from the proposed study may help society to better understand the role of obesity and gait movement modifications such as stride width changes on knee joint loading and functions during level and stair walk and help to decrease the risk of developing knee osteoarthritis.

**Confidentiality**

Your information will be kept confidential. Your research data and records will be stored securely and will be made available only to researchers who work on this study. The motion cameras will not record images of you. Your name will not be in any research data. Instead, a code number will replace your name on your data. Your name will not appear with the study results that will be presented at conferences and published in journals. Your data will be stored using password protected hard drives. Your data may be used for future research purposes after the completion of this study. If you decide to withdraw from the study, data collected up to that point may be used for research purposes, unless you request that it be destroyed.

**Contact Information**

If you have any questions about the study at any time or if you experience any problems as a result of participating in this study you can contact Derek Yocum or Dr. Songning Zhang at 1914 Andy Holt Ave. 136 HPER Bldg., The University of Tennessee and/or (865) 974-2091. Questions about your rights as a participant can be addressed to Compliance Officer in the Office of Research at the University of Tennessee at (865) 974-7697.

**Voluntary Participation and Withdrawal**

Your participation is entirely voluntary and your refusal to participate will involve no penalty or loss of benefits to which you are otherwise entitled. You may withdraw from the study at any time without penalty or loss of benefits to which you are otherwise entitled. Your participation in this study may be stopped by if you fail to follow the study procedures or if the principal investigator believes it is in your best interest to stop participation.

**Consent Statement**

I have read the above information. I agree to participate in this study. I have received a copy of this form.

Subject's Name: ________________________

Subject's Signature: __________________________ Date: _______________

Investigator's Signature: ______________________________ Date: _________________

**Appendix D: PAR-Q**

**PHYSICAL ACTIVITY READINESS QUESTIONNAIRE (PAR-Q)**

Regular physical activity is fun and healthy, and increasingly more people are starting to become more active every day. Being more active is very safe for most people. However, some people should check with their doctor before they start becoming much more physically active.

If you are planning to become much more physically active than you are now, start by answering the seven questions in the box below. If you are between the ages of 15 and 69, the PAR-Q will tell you if you should check with your doctor before you start. If you are over 69 years of age and you are not used to being very active, check with your doctor.

| No | Yes | |
|---|---|---|
| ☐ | ☐ | 1. Has your doctor ever said that you have a heart condition <u>and</u> that you should only do physical activity recommended by a doctor? |
| ☐ | ☐ | 2. Do you feel pain in your chest when you do physical activity? |
| ☐ | ☐ | 3. In the past month, have you had chest pain when you were not doing physical activity? |
| ☐ | ☐ | 4. Do you lose your balance because of dizziness or do you ever lose consciousness? |
| ☐ | ☐ | 5. Do you have a bone or joint problem that could be made worse by a change in your physical activity? |
| ☐ | ☐ | 6. Is your doctor currently prescribing drugs (for example water pills) for your blood pressure of heart condition? |
| | | 7. Do you know of <u>any other reason</u> why you should not do physical activity? |

Please note: If your health changes so that you then answer YES to any of these questions, tell your fitness or health professional. Ask whether you should change your physical activity plan.

**If you answered YES to one or more questions**

Talk to your doctor by phone or in person BEFORE you start becoming much more physically active of BEFORE you have a fitness appraisal. Tell your doctor about the PAR-Q and which questions you answered YES.

- You may be able to do any activity you want as long as you start slowly and build up gradually. Or you may need to restrict your activities to those which are safe for you. Talk to your doctor about the kinds of activities you wish to participate in and follow his/her advice.
- Find out which community programs are safe and helpful for you.

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**If you answered NO to all questions**

If you have answered NO honestly to all PAR-Q questions, you can be reasonably sure that you can:

- Start becoming much more physical active – begin slowly and build up gradually. This is the safest and easiest way to go.
- Take part if a fitness appraisal – this is an excellent way to determine your basic fitness so that you can plan the best way for you to live actively.

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Delay becoming much more active if:

- You are not feeling well because of a temporary illness such as a cold or a fever – wait until you feel better, or
  If you are or may be pregnant – talk to your doctor before you start becoming more active.

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I understand that my signature signifies that I have read and understand all the information on the questionnaire, that I have truthfully answered all the questions, and that any question/concerns I may have had have been addressed to my complete satisfaction.

__________________________________________________

Name (please print)

__________________________________________________

Signature

____________________

Date

**Appendix E: Demographic Questionnaire**

**Demographic Questionnaire**

Subject #:________________ Date (MM/DD/YY): _______/_____/_____

Age:_______________ Shoe Size (US):___________

Gender (circle one): Female Male

Any major lower extremity injuries of surgeries? (Circle One) Yes No

If yes, please explain further:

Injury:_____________________________________________________________

____________________________________________________________________________

_____________________________________________ Date:_______________

Any disorder affecting gait or balance? (Circle One) Yes No

Any lower extremity injuries within the past year? (Circle One) Yes No

If yes, please explain further:

Injury:_____________________________________________________________

__________________________________________________________ Date:_______________

Any pain while performing common activities of daily living, such as walking or using the stair?

(Circle One) Yes No

**Appendix F: Recruitment Flyer**

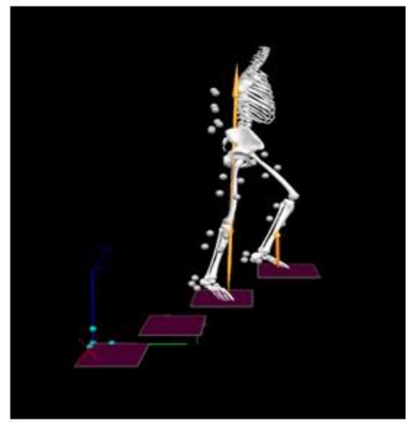
## RESEARCH PARTICIPANTS NEEDED FOR A STUDY ON THE EFFECTS OF BODY MASS AND STEP WIDTH DURING LEVEL WALKING AND STAIR NEGOTIATION

### Qualifications to participate in the study include:

- Between the ages of 18 and 35 yrs.
- Healthy with a BMI between 30-40.
- No major lower extremity injuries within the past year. No chronic condition affecting balance or leg function. Able to climb up and down stairs without use of handrail.

If you would like to participate or for more information contact Derek Yocum at the UT Biomechanics/Sports Medicine Lab.
Office: 865-974-2091
Email: dyocum@vols.utk.edu

Researchers from the Department of Kinesiology at UT are conducting research to understand the effects of body mass and step width during walking and stair climbing on the knee joint. Participants will attend one 2 hour testing session in the Biomechanics/Sports Medicine lab.



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**Appendix G: Individual Results for Selected Variables**

Table 9. Individual Peak Knee Extension Angle and Knee Extension ROM (deg): mean ± STD

| Subject | Peak Knee Extension Angle (deg) | | Knee Extension ROM (deg) | |
|---|---|---|---|---|
| | Preferred SW | Wide SW | Preferred SW | Wide SW |
| 2 | -10.224±2.874 | -11.463±2.391 | 53.717±2.984 | 49.870±2.686 |
| 4 | -8.513±1.267 | -12.595±1.608 | 53.231±1.727 | 49.702±2.323 |
| 5 | -11.769±2.323 | -14.767±2.773 | 52.695±2.402 | 49.244±3.653 |
| 9 | -2.970±1.898 | -17.559±0.691 | 66.550±2.095 | 64.488±5.782 |
| 10 | -10.890±4.814 | -12.107±2.204 | 55.917±4.947 | 54.422±3.063 |
| 11 | -9.938±1.684 | -4.701±3.410 | 58.007±3.422 | 51.541±2.827 |
| 12 | -7.321±1.306 | -6.912±1.995 | 57.434±2.673 | 56.889±3.632 |
| 13 | -14.498±1.828 | -18.976±3.024 | 55.440±1.170 | 47.037±2.921 |
| 14 | -10.542±1.560 | -12.581±1.871 | 60.907±1.730 | 59.752±1.506 |
| 15 | -9.640±0.961 | -6.776±2.213 | 57.247±3.235 | 58.587±0.771 |
| 16 | -10.907±2.203 | -14.033±3.610 | 64.544±3.033 | 56.431±5.083 |
| 17 | -8.599±1.331 | -8.309±1.849 | 55.004±1.405 | 50.992±2.632 |
| 18 | -3.263±1.413 | -5.253±1.007 | 63.973±1.959 | 58.547±1.352 |
| 19 | -15.034±1.511 | -16.948±1.716 | 51.657±2.631 | 46.486±1.873 |
| 20 | -12.844±2.583 | -14.548±2.026 | 58.979±3.005 | 58.863±7.586 |
| 21 | -0.607±0.786 | 1.623±1.781 | 62.901±2.450 | 64.978±2.595 |
| 22 | -1.643±3.750 | -5.302±2.831 | 56.421±3.698 | 54.327±5.127 |
| 23 | -10.118±2.091 | -8.229±3.122 | 44.194±1.621 | 41.399±1.696 |
| 24 | -3.915±2.623 | -7.074±1.428 | 55.693±3.020 | 54.444±1.676 |
| 25 | -12.595±1.608 | -10.784±1.911 | 55.561±1.698 | 52.012±4.114 |
| 26 | -14.767±2.773 | -11.278±1.843 | 62.094±3.751 | 51.173±5.467 |
| 27 | -17.559±0.691 | -19.124±1.919 | 54.596±1.264 | 47.844±2.326 |
| 28 | -12.107±2.204 | -11.261±1.626 | 42.356±2.242 | 40.003±1.817 |
| 29 | -4.701±3.410 | -3.696±1.861 | 65.102±3.015 | 66.256±3.927 |

Table 10. Peak Knee Abduction and Knee Abduction ROM (deg): mean ± STD

| Subject | Peak Knee Abduction Angle (deg) | | Knee Abduction ROM (deg) | |
|---|---|---|---|---|
| | Preferred SW | Wide SW | Preferred SW | Wide SW |
| 2 | 0.144±0.771 | 0.877±0.660 | -12.075±1.573 | -10.149±1.076 |
| 4 | -1.565±0.361 | -2.356±0.667 | -17.999±1.217 | -15.575±1.002 |
| 5 | -0.319±0.882 | -1.329±0.551 | -9.691±1.333 | -8.276±3.175 |
| 9 | -3.399±0.277 | -4.182±0.373 | -14.498±2.100 | -9.993±0.773 |
| 10 | -1.547±0.574 | -1.243±0.261 | -9.634±1.837 | -8.047±1.289 |
| 11 | -3.808±0.717 | -3.702±0.896 | -10.382±1.699 | -7.636±0.735 |
| 12 | -5.332±0.305 | -6.732±0.292 | -20.397±2.076 | -18.031±1.138 |
| 13 | -0.712±0.877 | 0.901±0.976 | -18.558±2.374 | -13.098±1.363 |
| 14 | 2.378±0.794 | 0.549±0.560 | -14.717±1.297 | -10.608±1.060 |
| 15 | -3.862±1.760 | -4.590±0.633 | -9.675±2.502 | -6.340±0.987 |
| 16 | 2.544±0.812 | 0.113±1.147 | -12.636±1.860 | -10.345±1.479 |
| 17 | -2.212±0.441 | -2.621±1.106 | -9.812±0.681 | -3.622±1.580 |
| 18 | -0.734±0.401 | -0.153±0.676 | -19.453±0.990 | -11.447±0.742 |
| 19 | 0.774±0.251 | -0.035±1.123 | -12.540±1.189 | -10.140±1.312 |
| 20 | -4.644±0.818 | -7.133±1.017 | -18.327±2.387 | -16.096±0.802 |
| 21 | -3.337±0.533 | -3.696±1.380 | -8.951±0.814 | -7.866±0.479 |
| 22 | -5.918±0.378 | -8.044±0.860 | -14.738±1.302 | -12.173±1.238 |
| 23 | -2.057±0.932 | -4.828±0.467 | -16.091±2.319 | -17.662±2.267 |
| 24 | -6.040±0.738 | -6.830±0.795 | -8.038±1.066 | -3.446±0.937 |
| 25 | -4.741±0.826 | -5.452±1.222 | -13.282±1.245 | -14.135±2.077 |
| 26 | -0.820±1.411 | -5.125±1.576 | -8.241±1.414 | -4.351±0.557 |
| 27 | -2.760±1.086 | -6.361±1.732 | -11.663±1.474 | -10.162±0.893 |
| 28 | -7.651±0.905 | -9.877±1.571 | -14.202±1.993 | -13.614±2.278 |
| 29 | -3.758±0.350 | -5.461±0.629 | -12.811±0.741 | -11.368±0.971 |

Table 11. Peak Knee Internal Rotation and Knee Internal Rotation ROM (deg): mean ± STD

| Subject | Peak Knee Internal Rotation Angle (deg) | | Knee Internal Rotation ROM (deg) | |
|---|---|---|---|---|
| | Preferred SW | Wide SW | Preferred SW | Wide SW |
| 2 | -5.469±1.843 | -6.886±0.669 | -9.870±2.610 | -11.609±1.441 |
| 4 | -11.629±2.777 | -12.057±1.957 | -11.973±2.924 | -10.528±2.775 |
| 5 | -11.057±4.555 | -11.034±4.250 | -6.615±3.480 | -7.678±4.714 |
| 9 | -10.143±2.228 | -11.038±1.840 | -11.585±1.484 | -14.948±1.592 |
| 10 | -7.441±0.999 | -8.281±2.445 | -5.596±1.723 | -7.943±2.949 |
| 11 | -7.774±2.108 | -9.068±2.284 | -11.282±2.676 | -13.886±2.828 |
| 12 | -13.663±1.856 | -14.070±0.863 | -12.381±1.558 | -13.131±1.038 |
| 13 | -15.723±2.145 | -12.933±2.540 | -21.038±2.075 | -17.174±2.462 |
| 14 | -8.532±2.616 | -4.742±1.764 | -9.984±2.397 | -8.269±2.414 |
| 15 | -15.673±0.853 | -13.980±0.500 | -6.520±1.560 | -8.683±1.557 |
| 16 | -7.348±2.119 | -3.226±3.216 | -11.430±2.161 | -7.882±3.312 |
| 17 | -6.543±0.597 | -10.386±0.772 | -6.545±1.394 | -12.950±1.675 |
| 18 | -8.347±1.196 | -8.075±4.395 | -4.621±1.120 | -3.987±2.531 |
| 19 | -12.651±0.927 | -11.465±1.867 | -11.170±2.066 | -15.001±2.577 |
| 20 | -12.551±4.766 | -10.323±3.414 | -8.868±3.285 | -17.971±3.819 |
| 21 | -10.684±1.227 | -12.936±4.174 | -4.699±1.525 | -13.751±2.737 |
| 22 | -13.530±2.509 | -13.581±2.346 | -11.236±3.577 | -13.779±3.696 |
| 23 | -11.751±1.579 | -13.077±1.514 | -16.632±1.475 | -18.723±1.910 |
| 24 | -9.217±0.939 | -7.358±1.346 | -7.786±1.358 | -10.368±1.780 |
| 25 | -5.363±2.118 | -4.003±1.734 | -2.031±0.897 | -3.601±1.562 |
| 26 | -12.428±2.108 | -8.642±4.316 | -11.471±2.429 | -9.536±5.912 |
| 27 | -6.329±2.528 | -3.063±1.861 | -6.158±3.301 | -4.446±2.667 |
| 28 | -10.706±1.874 | -4.360±2.856 | -7.675±1.426 | -7.991±2.274 |
| 29 | -2.551±0.998 | -3.622±1.647 | -2.723±1.398 | -2.788±1.715 |

Table 12. Loading-response and Push-off Peak Vertical GRF (BW): mean ± STD

| Subject | Loading-response Peak Vertical GRF (BW) | | Push-off Peak Vertical GRF (BW) | |
|---|---|---|---|---|
| | Preferred SW | Wide SW | Preferred SW | Wide SW |
| 2 | 1.079±0.045 | 1.115±0.022 | 1.102±0.028 | 1.121±0.028 |
| 4 | 1.107±0.031 | 1.058±0.042 | 1.241±0.036 | 1.088±0.129 |
| 5 | 1.221±0.040 | 1.188±0.040 | 1.318±0.070 | 1.250±0.054 |
| 9 | 1.084±0.050 | 1.059±0.038 | 1.245±0.048 | 1.231±0.057 |
| 10 | 1.249±0.032 | 1.236±0.052 | 1.328±0.031 | 1.332±0.070 |
| 11 | 1.204±0.045 | 1.118±0.021 | 1.298±0.108 | 1.233±0.104 |
| 12 | 1.126±0.033 | 1.072±0.041 | 1.203±0.053 | 1.228±0.057 |
| 13 | 1.137±0.089 | 1.095±0.040 | 1.396±0.037 | 1.535±0.071 |
| 14 | 1.095±0.025 | 1.070±0.038 | 1.183±0.050 | 1.217±0.039 |
| 15 | 1.217±0.064 | 1.248±0.018 | 1.197±0.049 | 1.250±0.094 |
| 16 | 1.233±0.026 | 1.140±0.055 | 1.299±0.045 | 1.249±0.039 |
| 17 | 1.194±0.042 | 1.164±0.019 | 1.522±0.028 | 1.356±0.091 |
| 18 | 1.101±0.013 | 1.017±0.016 | 1.273±0.072 | 1.269±0.065 |
| 19 | 1.100±0.022 | 1.066±0.042 | 1.222±0.028 | 1.242±0.067 |
| 20 | 1.119±0.034 | 0.972±0.017 | 1.150±0.038 | 1.331±0.070 |
| 21 | 1.165±0.042 | 1.119±0.012 | 1.228±0.045 | 1.218±0.038 |
| 22 | 1.068±0.020 | 1.168±0.035 | 1.213±0.043 | 1.222±0.078 |
| 23 | 1.134±0.048 | 1.180±0.022 | 1.230±0.032 | 1.194±0.073 |
| 24 | 1.302±0.054 | 1.180±0.026 | 0.991±0.027 | 1.232±0.048 |
| 25 | 1.080±0.043 | 1.001±0.047 | 1.097±0.060 | 1.021±0.039 |
| 26 | 1.115±0.032 | 1.068±0.069 | 1.179±0.077 | 1.115±0.078 |
| 27 | 1.157±0.032 | 1.136±0.052 | 1.175±0.133 | 1.346±0.129 |
| 28 | 1.040±0.013 | 1.050±0.020 | 1.110±0.092 | 1.168±0.111 |
| 29 | 1.053±0.017 | 1.067±0.026 | 1.100±0.049 | 1.106±0.067 |

Table 13. Loading-response and Push-off Peak Mediolateral GRF (BW): mean ± STD

| Subject | Loading-response Peak Vertical GRF (BW) | | Push-off Peak Vertical GRF (BW) | |
|---|---|---|---|---|
| | Preferred SW | Wide SW | Preferred SW | Wide SW |
| 2 | 1.079±0.045 | 0.127±0.007 | 0.062±0.009 | 0.127±0.016 |
| 4 | 1.107±0.031 | 0.184±0.013 | 0.086±0.011 | 0.167±0.026 |
| 5 | 1.221±0.040 | 0.171±0.022 | 0.049±0.014 | 0.127±0.011 |
| 9 | 1.084±0.050 | 0.125±0.011 | 0.025±0.006 | 0.120±0.018 |
| 10 | 1.249±0.032 | 0.146±0.020 | 0.041±0.012 | 0.113±0.023 |
| 11 | 1.204±0.045 | 0.101±0.012 | 0.014±0.019 | 0.061±0.010 |
| 12 | 1.126±0.033 | 0.158±0.010 | 0.061±0.013 | 0.148±0.019 |
| 13 | 1.137±0.089 | 0.120±0.010 | 0.037±0.008 | 0.128±0.021 |
| 14 | 1.095±0.025 | 0.133±0.013 | 0.048±0.006 | 0.132±0.010 |
| 15 | 1.217±0.064 | 0.190±0.016 | 0.050±0.018 | 0.133±0.013 |
| 16 | 1.233±0.026 | 0.173±0.013 | 0.062±0.009 | 0.151±0.007 |
| 17 | 1.194±0.042 | 0.123±0.016 | 0.026±0.014 | 0.112±0.023 |
| 18 | 1.101±0.013 | 0.133±0.006 | 0.075±0.015 | 0.159±0.005 |
| 19 | 1.100±0.022 | 0.156±0.015 | 0.038±0.004 | 0.133±0.023 |
| 20 | 1.119±0.034 | 0.123±0.009 | 0.030±0.009 | 0.150±0.008 |
| 21 | 1.165±0.042 | 0.125±0.005 | 0.059±0.007 | 0.131±0.009 |
| 22 | 1.068±0.020 | 0.088±0.026 | 0.015±0.009 | 0.076±0.014 |
| 23 | 1.134±0.048 | 0.124±0.009 | 0.030±0.013 | 0.092±0.019 |
| 24 | 1.302±0.054 | 0.109±0.007 | 0.013±0.009 | 0.087±0.017 |
| 25 | 1.080±0.043 | 0.145±0.009 | 0.074±0.014 | 0.142±0.015 |
| 26 | 1.115±0.032 | 0.217±0.022 | 0.084±0.017 | 0.194±0.012 |
| 27 | 1.157±0.032 | 0.114±0.011 | 0.029±0.009 | 0.123±0.019 |
| 28 | 1.040±0.013 | 0.162±0.018 | 0.066±0.012 | 0.188±0.028 |
| 29 | 1.053±0.017 | 0.121±0.004 | 0.038±0.005 | 0.098±0.003 |

Table 14. Peak Knee Extension Moment (Nm/kg): mean ± STD

| Subject | Peak Knee Extension Moment (Nm/kg) | |
|---|---|---|
| | Preferred SW | Wide SW |
| 2 | 1.726±0.084 | 1.759±0.067 |
| 4 | 1.418±0.064 | 1.548±0.111 |
| 5 | 1.527±0.155 | 1.369±0.229 |
| 9 | 1.186±0.065 | 1.196±0.072 |
| 10 | 1.854±0.065 | 1.887±0.198 |
| 11 | 1.502±0.109 | 1.222±0.066 |
| 12 | 1.495±0.086 | 1.608±0.137 |
| 13 | 1.387±0.099 | 1.406±0.082 |
| 14 | 1.721±0.103 | 1.897±0.088 |
| 15 | 1.519±0.116 | 1.756±0.095 |
| 16 | 1.571±0.067 | 1.602±0.188 |
| 17 | 1.415±0.076 | 1.560±0.043 |
| 18 | 1.594±0.045 | 1.474±0.136 |
| 19 | 1.337±0.067 | 1.379±0.079 |
| 20 | 1.394±0.117 | 1.329±0.071 |
| 21 | 1.832±0.065 | 1.939±0.127 |
| 22 | 1.564±0.052 | 1.539±0.088 |
| 23 | 1.392±0.136 | 1.494±0.116 |
| 24 | 2.052±0.143 | 1.887±0.090 |
| 25 | 1.534±0.194 | 1.807±0.246 |
| 26 | 1.640±0.044 | 1.634±0.144 |
| 27 | 1.633±0.072 | 1.670±0.123 |
| 28 | 1.367±0.047 | 1.363±0.050 |
| 29 | 1.499±0.055 | 1.618±0.061 |

Table 15. Loading-response Peak Knee Abduction Moment and Push-off Peak Knee Adduction Moment (Nm/kg): mean ± STD

| Subject | Loading-response Peak Knee Abduction Moment | | Push-off Peak Knee Adduction Moment | |
|---|---|---|---|---|
| | Preferred SW | Wide SW | Preferred SW | Wide SW |
| 2 | -0.398±0.039 | -0.372±0.040 | 0.164±0.018 | 0.160±0.060 |
| 4 | -0.413±0.054 | -0.382±0.090 | 0.173±0.037 | 0.322±0.114 |
| 5 | -0.321±0.075 | -0.397±0.109 | 0.242±0.033 | 0.360±0.061 |
| 9 | -0.143±0.035 | -0.272±0.034 | 0.226±0.045 | 0.319±0.058 |
| 10 | -0.281±0.049 | -0.360±0.055 | 0.249±0.023 | 0.315±0.030 |
| 11 | -0.326±0.035 | -0.317±0.056 | 0.205±0.026 | 0.219±0.025 |
| 12 | -0.362±0.038 | -0.249±0.063 | 0.142±0.008 | 0.327±0.061 |
| 13 | -0.428±0.066 | -0.414±0.020 | 0.130±0.033 | 0.179±0.025 |
| 14 | -0.489±0.071 | -0.368±0.012 | 0.109±0.037 | 0.127±0.017 |
| 15 | -0.195±0.060 | -0.345±0.041 | 0.173±0.052 | 0.162±0.096 |
| 16 | -0.303±0.053 | -0.206±0.039 | -0.010±0.023 | 0.066±0.026 |
| 17 | -0.220±0.053 | -0.217±0.033 | 0.191±0.029 | 0.268±0.043 |
| 18 | -0.540±0.046 | -0.442±0.049 | 0.187±0.055 | 0.194±0.038 |
| 19 | -0.476±0.059 | -0.383±0.041 | 0.121±0.019 | 0.225±0.069 |
| 20 | -0.184±0.031 | -0.054±0.020 | 0.373±0.037 | 0.391±0.029 |
| 21 | -0.058±0.041 | -0.075±0.029 | 0.105±0.023 | 0.145±0.038 |
| 22 | -0.246±0.043 | -0.238±0.047 | 0.326±0.027 | 0.360±0.068 |
| 23 | -0.369±0.054 | -0.429±0.051 | 0.231±0.060 | 0.353±0.069 |
| 24 | -0.331±0.041 | -0.223±0.046 | 0.196±0.030 | 0.304±0.032 |
| 25 | -0.092±0.041 | -0.106±0.053 | 0.116±0.023 | 0.278±0.052 |
| 26 | -0.275±0.046 | -0.121±0.072 | -0.144±0.035 | 0.094±0.095 |
| 27 | -0.166±0.057 | -0.101±0.037 | 0.068±0.024 | 0.206±0.069 |
| 28 | -0.202±0.039 | -0.041±0.036 | 0.371±0.042 | 0.470±0.044 |
| 29 | -0.188±0.049 | -0.116±0.016 | 0.096±0.016 | 0.129±0.023 |

Table 16. Peak Loading-response Peak Knee External Rotation Moment and Push-off Peak Knee Internal Rotation Moment (Nm/kg): mean ± STD

| Subject | LR Peak Knee External Rotation Moment (Nm/kg) | | PO Peak Knee Internal Rotation Moment (Nm/kg) | |
|---|---|---|---|---|
| | Preferred SW | Wide SW | Preferred SW | Wide SW |
| 2 | 0.213±0.029 | 0.224±0.028 | -0.123±0.022 | -0.102±0.027 |
| 4 | 0.083±0.029 | 0.181±0.068 | -0.135±0.010 | -0.104±0.019 |
| 5 | 0.154±0.037 | 0.193±0.039 | -0.118±0.034 | -0.111±0.021 |
| 9 | 0.041±0.013 | 0.137±0.035 | -0.175±0.016 | -0.123±0.025 |
| 10 | 0.149±0.017 | 0.208±0.055 | -0.143±0.039 | -0.123±0.004 |
| 11 | 0.057±0.010 | 0.119±0.031 | -0.118±0.014 | -0.087±0.014 |
| 12 | 0.160±0.014 | 0.209±0.029 | -0.097±0.021 | -0.089±0.021 |
| 13 | 0.114±0.041 | 0.187±0.034 | -0.191±0.014 | -0.124±0.031 |
| 14 | 0.140±0.021 | 0.190±0.006 | -0.101±0.012 | -0.087±0.007 |
| 15 | 0.164±0.027 | 0.381±0.055 | -0.153±0.031 | -0.156±0.040 |
| 16 | 0.058±0.031 | 0.090±0.023 | -0.143±0.039 | -0.099±0.012 |
| 17 | 0.115±0.027 | 0.215±0.028 | -0.144±0.014 | -0.087±0.013 |
| 18 | 0.214±0.029 | 0.178±0.045 | -0.134±0.024 | -0.105±0.017 |
| 19 | 0.175±0.017 | 0.198±0.011 | -0.099±0.011 | -0.096±0.015 |
| 20 | 0.067±0.028 | 0.113±0.024 | -0.106±0.007 | -0.059±0.009 |
| 21 | 0.123±0.018 | 0.122±0.032 | -0.106±0.021 | -0.114±0.014 |
| 22 | 0.086±0.026 | 0.179±0.038 | -0.164±0.019 | -0.175±0.046 |
| 23 | 0.109±0.044 | 0.162±0.031 | -0.147±0.015 | -0.102±0.025 |
| 24 | 0.303±0.025 | 0.272±0.027 | -0.125±0.013 | -0.115±0.022 |
| 25 | 0.146±0.059 | 0.177±0.034 | -0.077±0.012 | -0.098±0.029 |
| 26 | 0.272±0.029 | 0.229±0.062 | -0.064±0.017 | -0.098±0.018 |
| 27 | 0.220±0.011 | 0.262±0.051 | -0.081±0.007 | -0.056±0.015 |
| 28 | 0.153±0.011 | 0.159±0.019 | -0.024±0.020 | -0.020±0.019 |
| 29 | 0.117±0.019 | 0.139±0.010 | -0.118±0.004 | -0.121±0.021 |

## Appendix H: Condition Order and Results from Statistical Tests

Table 17. Random Numbers indicating condition order: 1- Condition Performed First

| Subject Number | Preferred SW | Wide SW | Random Number (0-1) |
|---|---|---|---|
| 2 | 2 | 1 | 0.815 |
| 4 | 2 | 1 | 0.906 |
| 5 | 1 | 2 | 0.127 |
| 9 | 2 | 1 | 0.913 |
| 10 | 2 | 1 | 0.632 |
| 11 | 1 | 2 | 0.098 |
| 12 | 1 | 2 | 0.279 |
| 13 | 1 | 2 | 0.371 |
| 14 | 1 | 2 | 0.158 |
| 15 | 2 | 1 | 0.971 |
| 16 | 1 | 2 | 0.485 |
| 17 | 2 | 1 | 0.800 |
| 18 | 1 | 2 | 0.142 |
| 19 | 1 | 2 | 0.422 |
| 20 | 2 | 1 | 0.916 |
| 21 | 1 | 2 | 0.401 |
| 22 | 2 | 1 | 0.656 |
| 23 | 1 | 2 | 0.036 |
| 24 | 1 | 2 | 0.097 |
| 25 | 2 | 1 | 0.758 |
| 26 | 1 | 2 | 0.392 |
| 27 | 1 | 2 | 0.171 |
| 28 | 2 | 1 | 0.706 |
| 29 | 1 | 2 | 0.277 |

Table 18. Results from Independent T-tests for Age, Height, Leg Length, Body Fat Percent, and BMI

| | Levene's Test for Equality of Variances | | t-test | |
|---|---|---|---|---|
| | F | p | t | p |
| Age | 17.851 | 0.000 | -2.209 | 0.054 |
| Height | 3.861 | 0.062 | -1.242 | 0.227 |
| Leg Length | 6.919 | 0.015 | -1.274 | 0.217 |
| Body Fat % | 0.336 | 0.568 | -4.763 | 0.000 |
| BMI | 4.445 | 0.047 | -10.516 | 0.000 |

Table 19. Results of F tests, p, and $\eta^2$ values for selected ANOVAs

| Variable | Main Effect (Preferred vs Wide) F (1,22) | p | $\eta^2$ | Interaction F (1,22) | p | $\eta^2$ |
|---|---|---|---|---|---|---|
| Step Width | 564.820 | 0.000 | 0.963 | 2.574 | 0.123 | 0.105 |
| LR ML GRF | 539.609 | 0.000 | 0.961 | 17.707 | 0.000 | 0.446 |
| PO ML GRF | 367.437 | 0.000 | 0.944 | 18.211 | 0.000 | 0.453 |
| LR Vertical GRF | 4.525 | 0.045 | 0.171 | 1.146 | 0.296 | 0.050 |
| PO Vertical GRF | 0.536 | 0.472 | 0.024 | 0.914 | 0.349 | 0.040 |
| LR Peak Knee Extension Moment | 3.133 | 0.091 | 0.125 | 1.316 | 0.264 | 0.056 |
| LR Peak KAbM | 8.110 | 0.009 | 0.269 | 4.270 | 0.051 | 0.163 |
| PO Peak KAdM | 39.296 | 0.000 | 0.641 | 6.094 | 0.022 | 0.217 |
| LR Peak Knee External Rotation Moment | 17.215 | 0.000 | 0.439 | 0.094 | 0.763 | 0.004 |
| PO Peak Knee Internal Rotation Moment | 8.631 | 0.008 | 0.282 | 1.348 | 0.258 | 0.058 |
| Peak Knee Extension Angle | 0.495 | 0.489 | 0.022 | 0.953 | 0.340 | 0.041 |
| Peak Knee Abduction Angle | 27.515 | 0.000 | 0.556 | 9.723 | 0.005 | 0.307 |
| Peak Knee Internal Rotation Angle | 2.933 | 0.101 | 0.118 | 0.688 | 0.416 | 0.030 |
| Knee Extension ROM | 22.713 | 0.000 | 0.508 | 0.018 | 0.895 | 0.001 |
| Knee Abduction ROM | 41.966 | 0.000 | 0.656 | 7.958 | 0.010 | 0.266 |
| Knee Internal Rotation ROM | 6.059 | 0.022 | 0.216 | 0.553 | 0.465 | 0.025 |

Note: LR: Loading-Response, PO- Push-off Response, ML-mediolateral, GRF-Ground Reaction Force, ROM-Range of Motion

## Vita

Derek Yocum was born in Wabash, IN in 1990 to Denise and Dewayne Yocum. He is the youngest of two children and grew up in Gilead, IN. Derek graduated from North Miami High School in 2009. He first attended the Huntington University, and later graduated from the Ball State University with a degree in Exercise Science. After graduation, he completed a M.S. in Kinesiology with an emphasis in Biomechanics at the University of Tennessee in Fall 2016.